

2002

Cumulative loading of the lumbar spine during non-occupational activities.

Christa L. Lauder
University of Windsor

Follow this and additional works at: <http://scholar.uwindsor.ca/etd>

Recommended Citation

Lauder, Christa L., "Cumulative loading of the lumbar spine during non-occupational activities." (2002). *Electronic Theses and Dissertations*. Paper 1597.

This online database contains the full-text of PhD dissertations and Masters' theses of University of Windsor students from 1954 forward. These documents are made available for personal study and research purposes only, in accordance with the Canadian Copyright Act and the Creative Commons license—CC BY-NC-ND (Attribution, Non-Commercial, No Derivative Works). Under this license, works must always be attributed to the copyright holder (original author), cannot be used for any commercial purposes, and may not be altered. Any other use would require the permission of the copyright holder. Students may inquire about withdrawing their dissertation and/or thesis from this database. For additional inquiries, please contact the repository administrator via email (scholarship@uwindsor.ca) or by telephone at 519-253-3000ext. 3208.

INFORMATION TO USERS

This manuscript has been reproduced from the microfilm master. UMI films the text directly from the original or copy submitted. Thus, some thesis and dissertation copies are in typewriter face, while others may be from any type of computer printer.

The quality of this reproduction is dependent upon the quality of the copy submitted. Broken or indistinct print, colored or poor quality illustrations and photographs, print bleedthrough, substandard margins, and improper alignment can adversely affect reproduction.

In the unlikely event that the author did not send UMI a complete manuscript and there are missing pages, these will be noted. Also, if unauthorized copyright material had to be removed, a note will indicate the deletion.

Oversize materials (e.g., maps, drawings, charts) are reproduced by sectioning the original, beginning at the upper left-hand corner and continuing from left to right in equal sections with small overlaps.

**ProQuest Information and Learning
300 North Zeeb Road, Ann Arbor, MI 48106-1346 USA
800-521-0600**

UMI[®]

Cumulative Loading of the Lumbar Spine During Non-Occupational Activities

By

Christa L. Lauder

A Thesis

**Submitted to the Faculty of Graduate Studies and Research
Through Human Kinetics**

**In Partial Fulfillment of the Requirements of
The Degree of Master of Human Kinetics at the
University of Windsor**

Windsor, Ontario, Canada

2002

© 2002 Christa L. Lauder



**National Library
of Canada**

**Acquisitions and
Bibliographic Services**

**385 Wellington Street
Ottawa ON K1A 0N4
Canada**

**Bibliothèque nationale
du Canada**

**Acquisitions et
services bibliographiques**

**385, rue Wellington
Ottawa ON K1A 0N4
Canada**

Your file Votre référence

Our file Notre référence

The author has granted a non-exclusive licence allowing the National Library of Canada to reproduce, loan, distribute or sell copies of this thesis in microform, paper or electronic formats.

The author retains ownership of the copyright in this thesis. Neither the thesis nor substantial extracts from it may be printed or otherwise reproduced without the author's permission.

L'auteur a accordé une licence non exclusive permettant à la Bibliothèque nationale du Canada de reproduire, prêter, distribuer ou vendre des copies de cette thèse sous la forme de microfiche/film, de reproduction sur papier ou sur format électronique.

L'auteur conserve la propriété du droit d'auteur qui protège cette thèse. Ni la thèse ni des extraits substantiels de celle-ci ne doivent être imprimés ou autrement reproduits sans son autorisation.

0-612-75839-7

Canada

Abstract

CUMULATIVE LOADING OF THE LUMBAR SPINE DURING NON- OCCUPATIONAL ACTIVITIES

The purpose of the study is to document peak and cumulative estimates on the lumbar spine in the laboratory setting as well as cumulative loading estimates on the lumbar spine in a field setting. Further, this study was also designed to compare video methods and electromyography methods of estimating cumulative compression on the lumbar spine. The variables measured in the laboratory study include both peak and cumulative estimates of compression, reaction shear, joint shear and moment, while the field study measured only cumulative estimates of these variables.

Results indicated that there is a significant amount of low back loading that occurs within the home while performing non-occupational activities. For the purpose of this study, non-occupational refers to tasks performed within the home. Cumulative estimates of compression, reaction shear, joint shear and moment measured over a two-hour duration for five subjects in the field setting are equal to almost half the cumulative estimates of the same variables documented by Norman et al (1999) in an industrial setting over the duration of an eight-hour work day. Further, three of the seven laboratory tasks yielded peak compression estimates exceeding Mitals' female tolerance limit of 2700 N for the lumbar spine. None of the subjects exhibited peak loading estimates in excess of the NIOSH allowed limit of 3400 N at the lumbar spine. Statistical analysis revealed that the Compression Normalized EMG technique (CNEMG) developed by Potvin et al (1990), is an accurate method of estimating cumulative

compression at the lumbar spine for the tasks studied when compared against video methods. A relative error of less than 10% between methods was achieved for all tasks but sitting which had an average relative error of 18%.

The majority of tasks in both the lab and field portion of the study required the subjects to handle loads less than 10 kg. Further, peak estimates of lumbar compression reported in the literature for industrial tasks still display higher loading profiles. This gives the impression that household activities are less physically challenging than industrial tasks. This assumption is misleading as only a small number of tasks and subjects have been studied outside the industrial setting thus far. The spine is loaded in a number of ways during the day and further analysis with a greater variety of asymmetric tasks will undoubtedly yield greater loading estimates for non-occupational activities.

Acknowledgments

I would like to thank all of my subjects, especially those participating in the field portion of the study as they allowed me into their homes to participate in a lengthy collection session. I would also like to thank Monica Haumann for all her time and help during this project. To the members of my thesis committee, Dr. Dave Andrews, Dr. Jim Potvin, Dr. Jack Callaghan and Dr. Anne Snowden, thank-you for your time and insight. Dr. David Andrews thank-you for your guidance and patience over the duration of this lengthy project. Dr. Jim Potvin, thank-you for your advice as well as the various opportunities you have provided me with during my time at the University of Windsor. To my mom Mary Lauder for her support and my dad Harry Lauder for his technical advice, thank-you. Special thanks to my brother Ian Lauder, my sister Laura Lauder, my aunt Liz Vanderkwaak as well as my friends (especially Holly Gaylord and Erin Kepran) who were integral in the completion of this project. I would like to dedicate this thesis to my grandmothers Eleanor Lauder and the Late Catherine McKeown.

Table of Contents

Abstract	iii
Acknowledgments	v
List of Figures	viii
List of Tables	ix
Introduction	1
Statement of Purpose	2
Statement of Hypotheses	2
Review of Literature	7
Tissues of the Lumbar Spine	7
Muscle	7
Vertebrae	8
Intervertebral Discs	9
Ligaments	10
Lumbar Spine Kinematics	11
Flexion	11
Extension	12
Lateral Bend	12
Axial Rotation	13
Lumbar Spine Kinetics	13
Tissue Failure	13
Moment	14
Compression	15
Shear	17
Flexion	18
Lateral Bend	20
Torsion	21
Cumulative Loading	22
Fatigue Failure	22
Cumulative Loading and Low Back Pain	23
Proposed Injury Mechanisms	24
Non-Occupational Activities	27
Exposure Limits	29
Peak Loading	29
Compression Tolerance Limits	29
Methods for Quantifying Cumulative Loads	30
Comparison of Methods	32

Methodology	34
Video Methods	34
Compression Normalized EMG Method	34
EMG Normalization	35
Data Acquisition	36
Field Collection	36
Laboratory Collection	40
Subjects	42
Data Treatment	42
Video Data	42
EMG Data	43
Submaximal Voluntary Contraction Session	44
Dependant Variables	44
Field Study	44
Laboratory Study	44
Data Analysis	45
 Results	 47
Field Study	47
Task Distributions	47
Cumulative Estimates of Compression, Moment and Shear Forces	49
Video Techniques vs CNEMG Methods	50
Laboratory Study	51
Peak Loading Estimates	51
Cumulative Loading Estimates	52
Video Techniques vs CNEMG Methods	53
 Discussion	 57
Field Study	57
Laboratory Study	65
Hypotheses	67
 Conclusions	 71
Field Study	71
Laboratory Study	72
 References	 77
 Vita Auctoris	 77

List of Figures

		Page
Figure 1	Average task distribution for all subjects in the field study.	47
Figure 2	Amplitude probability distribution function for the field collection session using the CNEMG technique, subject 2.	49
Figure 3	Peak estimates of lumbar compression versus tolerance limits for the basket, folding and garbage tasks.	52
Figure 4	Force time history for CNEMG and video analysis techniques of estimating lumbar compression for the folding task, subject 1.	54
Figure 5	Amplitude probability distribution function for CNEMG and video analysis techniques, shelves task, subject 9.	55
Figure 6	Linear regression plot for the basket task.	56

List of Tables

		Page
Table 1	Definitions of tasks performed during field portion of study.	37
Table 2	Description of tasks performed by subjects during the laboratory portion of the study.	41
Table 3	Percentage of time spent performing each task in the field study for each task and subject.	48
Table 4	Means and standard deviations for cumulative estimates of compression, reaction shear, joint shear and moment for each subject in the field study using the video method.	49
Table 5	Cumulative estimates of compression for all subjects in the field study.	50
Table 6	Means and standard deviations for peak estimates of compression reaction shear, joint shear and moment for all tasks and subjects in the laboratory study using video techniques.	51
Table 7	Means and standard deviations for cumulative estimates of compression, reaction shear, joint shear and moment for all tasks and subjects in the laboratory study using video techniques.	53
Table 8	Mean values and standard deviations for relative error between cumulative estimates via video analysis and CNEMG techniques for all tasks.	54
Table 9	Summary of significant findings for CNEMG and video techniques of estimating cumulative compression at L4/L5.	56

CHAPTER I

1.0 INTRODUCTION

Low back pain (LBP) is a multivariate phenomena affecting millions worldwide and is often the result of biomechanical stresses placed on the various structures of the vertebral column. Degeneration of the disc, however, is thought to be the major factor for low back pain and injury as disc herniation, segmental instability and the majority of lumbar syndromes originate from this phenomena (Vanharanta, 1989). In recent years, low back pain, has become a major health concern due to its prevalence, cost and associated pain and suffering. Fymoyer and Pope (1983) have reported that the lifetime prevalence of low back pain ranges from 60-90%, with an annual incidence of 5%. Further, it is estimated that \$24 billion is spent per year for direct medical cost of managing low back pain (Frymoyer & Cats-Baril, 1991).

There are several biomechanical stressors that place the lumbar spine at risk for injury. Compression has been shown to cause fractures in the vertebral endplate in response to peak loads ranging from 3-12 kN (Timm & Malone, 1989) as well as the intervertebral discs in response to peak loads ranging from 2.8-13 kN when combined with lateral or forward bending (Adams & Dolan, 1995). Shear loading has been shown to result in failure of the pars interarticularis, vertebral endplate (Yingling et al, 1999) as well as the facets (Adams & Hutton, 1981). Fractures of the pars interarticularis occur in response to peak loads of approximately 2000 N (Troup, 1976). Further, tension loading acts to increase the risk of strains and tears to the passive tissues of the joints such as the

ligaments and intervertebral discs (Myklebust et al, 1986). Damage of the ligament can occur when the bending moment acting on the spine is about 60 Nm while complete failure occurs at 140-185 Nm (Adams & Dolan, 1995). Trunk postures deviating from anatomically neutral positions, such as mild or severe flexion and twisting or lateral bending place the lumbar spine under the influence of one or more of these biomechanical stressors, therefore, increasing the risk of injury (Punnet et al, 1991). Trunk velocity and extensor moment have also been shown to be related to an increased risk of injury to the low back (Marras et al. 1993).

Repeated or constant application of biomechanical stressors or cumulative loading of the lumbar spine has also been identified as a risk factor for injury (Kumar, 1990; Norman et al, 1998). Institutional aides with pain were reported to have a significantly higher daily cumulative compression loading profile than those without pain (Kumar, 1990), while auto workers with pain displayed a significantly higher daily cumulative loading profile for lumbar moment and usual hand force than those without pain (Norman et al, 1998). Video based methods of estimating cumulative loading profiles were used in both studies whereby joint coordinate data obtained from frames of video were input to a biomechanical model in an effort to obtain loading estimates of shear and compression. These values were extrapolated and a daily dose of shear and compression was estimated. EMG techniques to estimate daily cumulative loads have also been developed, whereby EMG data is multiplied by a scale factor to estimate compression loading throughout the entire cycle, and is termed compression normalized EMG (Potvin et al, 1990).

It is believed that repeated application of a load to biological tissue results in accumulated microtrauma, thereby reducing its ability to withstand subsequent loads (Whiting & Zernicke, 1998). As a result failure occurs at much lower loads when loaded repeatedly (Brinckmann et al., 1989; Adams & Dolan, 1984). The strength of a tissue when loaded cyclically is thus a function of both the number of loading cycles (Brinckmann et al., 1989), and the magnitude of the load (Brinckmann et al., 1989; Hansson et al., 1987).

There may be several mechanisms through which cumulative loading results in injury and may be dependent on loading characteristics such as the plane, magnitude and duration of loading. Generally, LBP and injury is the result of the responses of either the passive tissues or musculature of the lumbar spine to cumulative loads. Cyclic flexion loading of the ligamentous spine has been shown to increase the laxity of the passive tissues, thereby exposing the spine to instability and thus an increased risk for injury (Goel et al., 1988). Further, cyclic passive loading of the spine resulting in laxity of the passive tissues has also been shown to reduce the muscle activity as a result of the desensitization of mechanoreceptors which act to monitor muscular stabilizing forces within the joint (Solomonow et al., 1999).

During prolonged cyclic loading muscles of the lumbar spine can become fatigued and thus the force they are able to produce is reduced (Potvin & Norman, 1993). This may result in more demand on the passive structures of the spine placing them at a

greater risk of injury. Further, fatigue in the agonist muscles is often accompanied by increased activity of the antagonists in an effort to stabilize the joint (Potvin & O' Brien, 1998). This increased muscle activity has been shown to increase the joint compressive penalty, and thus an increased risk of injury (McGill, 1992; Granata & Marras, 1995).

Much of the biomechanics research done in the area of low back pain and injury has been done with the intent of quantifying the risk of injury and developing exposure limits for safe working conditions in the workplace. Typically, peak loading conditions resulting in failure of the tissues of the lumbar spine has been the focus. Peak levels of compression in the range of 3-12 kN have been reported to result in failure of the tissues of the lumbar spine (Timm & Malone, 1989). These values, combined with epidemiologic studies identifying a dose response link between compression loading and injury to the spinal tissue (Herrin et al., 1986; Bringham & Garg, 1983) have been used to establish compression tolerance limits for manual materials handling tasks (examples: the National Institute for Occupational Safety and Health (1981) Action Limit and Maximum Permissible Limit of 3400 N and 6400 N, respectively, as well the limits proposed by Mital et al (1993) of 3900 N for males and 2700 N for females.

Cyclic compression loading of the lumbar spine has resulted in failure at loads as low as 2240 N, which falls below any of the reported peak tolerance limits (Adams & Dolan, 1985). It is necessary, therefore, to develop cumulative exposure limits to be used in conjunction with peak tolerance limits in order to reduce the incidence of low back

pain and injury during manual materials handling tasks. Before this can be done certain tasks must be completed:

- 1) A common method of measuring cumulative load that is both practical and results in minimal error in the estimation of cumulative loading profiles must be identified. Callaghan et al. (2001), assessed the error present amongst five video-based techniques. Rectangular integration at a reduced sampling rate (5 Hz), was the only method that did not result in significant error when compared against the gold standard of 30 Hz (Callaghan et al., 2001). Video analysis requires a substantial amount of time and effort, even when processing at 5 Hz. EMG methods of estimating cumulative compression profiles may, therefore, be a more practical approach in some situations. Potvin et al. (1990) performed a study to compare EMG and video techniques of estimating cumulative lumbar compression during lifting tasks. To date these techniques have not been compared for prolonged loading cycles encompassing a wide variety of tasks.**
- 2) Loading profiles during both work and non-occupational activities must be quantified. Loading of the spine occurs continuously throughout the day from the time an individual wakes up until the time they go to bed. If it is accepted that most injury is the result of cumulative microtrauma placed on the body's tissues, then all loading conditions encountered throughout the day must be considered if spinal injury is to be fully understood. These include loads encountered at work, in the home as well as leisure or recreational activities. Typically, low back biomechanics research has focused on loading at work, and to the best of the**

authors' knowledge there has been little quantification of loading profiles during non-occupational activities such as housekeeping and childcare.

- 3) A dose response link between level of loading exposure and an increased risk of injury must be identified.**

1.1 STATEMENT OF PURPOSE

The purpose of this thesis is twofold:

- 1) The first objective is to document peak and cumulative estimates for compression, shear and moment during non-occupational activities in the laboratory and cumulative estimates of the same variables in the field.**
- 2) The second objective was to compare video-based versus EMG methods of estimating cumulative compression profiles in the laboratory and the field.**

1.2 STATEMENT OF HYPOTHESIS

- 1) Peak low back spinal loads estimated during non-occupational activities will be similar in magnitude to those reported for many work activities involving manual materials handling tasks.**
- 2) Estimates of cumulative compression will be lower for non-occupational activities over an eight-hour day than for industrial work.**
- 3) Low back cumulative compressive loads determined by video (sagittal view of the right side of the body) and EMG methods will not be significantly different from each other for two-dimensional symmetric tasks.**

CHAPTER II

2.0 REVIEW OF LITERATURE

2.1 TISSUES OF THE LUMBAR SPINE

The human body acts as a lever system and its ability to bear loads is dependant upon the interaction of its various tissues. With respect to the vertebral column, several distinct structures including joints, ligaments, muscles and bones act together to allow mechanical functioning of the spine.

2.1.1 Muscle

Muscle tissue is composed of specialized fibers capable of generating force through the contraction process. Skeletal muscle, therefore, acts to generate movement about the joints as well as maintain stability of the trunk. With respect to the lumbar spine, eleven major muscle groups are required to regulate the activities of the trunk, including the psoas, quadratus lumborum, intersegmental complex (interspinales and intertransversarii muscle groups), multifidus, rotators, erector spinae, rectus and transversus abdominis, external and internal obliques and the latissimus dorsi (Timm & Malone, 1989).

The muscles of the trunk bear the majority of the load placed on the spine and are responsible for most of the restorative moment during bending activities (McGill & Norman, 1985; Potvin et al., 1991). This muscle activity acts to compress vertebral joints of the spine and thus has implications for injury. Further, fatigue of these structures, and thus a reduction in force generating capacity may result in a greater demand placed on the

passive structures of the spine such as the ligaments or intervertebral discs, placing them at a greater risk for injury.

2.1.2 Vertebrae

The vertebrae are the bony portions of the spine and are composed of a body, a pedicle and a neural arch. The laminae protrude backwards from the pedicle to form the neural arch, and possess several bony protrusions known as the vertebral processes. These structures act to protect the spinal chord, resist compressive forces and provide a bony interface for the muscles and ligaments of the spine.

The lumbar vertebrae, found in the low back are the largest and strongest as the amount of load that must be sustained is greater at the inferior end of the spine due to the weight of the body segments above it. In fact, the lumbar vertebral bodies together with the intervertebral discs act to resist 80% of compressive forces placed on the spine in an upright standing posture (Adams & Dolan, 1995). Depending on posture the remaining 20% falls upon the apophyseal joints (Adams & Dolan, 1995).

With respect to the lumbar spine, there are two distinct joint types. The facet or apophyseal joint is formed by the junction of superior vertebral articular processes with the inferior vertebral articular processes of adjacent vertebrae. These joints act to restrict the range of motion of the lumbar spine via a “doorstop mechanism”, whereby corresponding processes are pressed against one another. Apophyseal joints help to resist compression but are best able to resist horizontal forces and thus limit the range of axial

rotation. Further they are capable of resisting maximum forward shearing forces approximating 2 kN (Cyron et al., 1976). The other type of joint found in the lumbar spine is between the vertebral bodies, structures separated by the intervertebral disc.

2.1.3 Intervertebral Discs

Intervertebral discs are comprised of two main structures. The annulus fibrosus is an outer fibrous ring that acts as a tensile shell for the disc and blends with the layer of cartilage found on the endplate of the superior and inferior vertebrae. Collagen fibers of the annulus are arranged in 10 to 12 lamellar sheets whose directions oppose each other at angles of approximately 30 degrees (Whiting & Zernicke, 1998).

The nucleus pulposus is an inner gel-like structure with a 70% to 90% water content. The remaining portion is composed of ground substance, collagen and chondrocytes. As a result of its fluid composition, the nucleus allows the disc to deform when it is loaded, thereby allowing motion of the spinal segments. Further, the forces received by the disc during loading are transmitted equally in all directions throughout the disc and are countered by the tensile strength of the annulus. In this manner, loading forces are transmitted to the vertebral bodies (Watkins, 1999). Consequently, the main functions of the disc are to allow movement of the spine, absorb shock transfer between adjacent vertebrae, and aid in the resistance of loads placed on the spine such as compression shear, and torsion.

The disc is well protected from the effects of extreme postures by other structures within the spinal column such as the posterior ligaments and the apophyseal joints of the

vertebrae. As a result, damage to the disc is believed to be secondary to strains placed on these protective structures. Failure of the disc can occur in response to a single loading condition with forces ranging from 2.8-13 kN when lateral or forward bending is present, or in response to cyclic loads of approximately 3 kN (Adams & Dolan, 1995). Further, Callaghan and McGill (2001) reported herniation of porcine discs with modest compression forces but highly repetitive flexion/extension moments.

2.1.4 Ligaments

Ligaments are the fibrous structures, which attach bone to bone, thereby stabilizing the joint and restricting its range of motion. These structures are composed mainly of collagen and therefore possess a high tensile strength but will buckle under shear loading (LeVeau, 1991). According to Adams and Hutton (1980) and Adams and Hutton (1991) the first sign of damage to a ligament occurs in response to a bending moment of approximately 60 Nm with gross damage at 120 Nm (Nuemann et al., 1991) and complete failure at 140 to 185 Nm (Osvalder et al., 1990; Osvalder et al., 1993) with a joint-flexion angle ranging from 5 to 20 degrees. Myklebust et al. (1986), reported failure loads at the lumbar level ranging from 38 N for the posterior longitudinal ligament corresponding to a 7 mm deflection. Further, this study found a progressive increase in strength of spinal ligaments from the cervical vertebrae to the lumbar vertebrae, a finding that is consistent with the increased strength of the lumbar vertebrae compared to the vertebrae above (Myklebust et al., 1986).

Further, due to their viscoelastic properties ligaments' resistance to flexion increases under rapid loading conditions. However, sustained flexion decreased resistance 42% in 5 min and by 67% in an hour (Adams & Dolan, 1995). Rapid flexion is therefore more likely to cause failure in the ligament while prolonged flexion is more likely to cause failure at the ligament attachment site to the bone (avulsion fracture).

2.2 LUMBAR SPINE KINEMATICS

Range of motion (ROM) refers to the planes and distances through which a joint is free to operate. With respect to the vertebral column, the intervertebral joints (IV joints) allow for motion in the sagittal, frontal and transverse planes, which corresponds to flexion/extension, lateral bend, and axial rotation or twist, respectively. In general, the functional ROM of the IV joint is determined by the combined effects of the shape and orientation of the vertebrae, ligaments, thickness of the IV disc as well as the surrounding musculature. It is believed that the ROM is linked to the stability of the joint, and motion beyond the limits described in the following paragraphs increases the risk of injury to the relevant tissues (Whiting & Zernicke, 1998). Further, trunk postures, which simply deviate from anatomically neutral, may have implications for injury (Punnett et al, 1991).

2.2.1 Flexion

Flexion can be defined as the decrease in angle of the trunk and the pelvis in the sagittal plane. Motion of the spine in this direction is limited by the supraspinous ligament, interspinous ligaments, intertransverse ligaments, posterior longitudinal

ligament, capsules of the facet joints and posterior aspects of the IV disc (Kapandji, 1974). Flexion at each IV joint in the lumbar spine has been reported to range from 1-24 degrees with greater values found at more inferior aspects of the spine. The flexion limit of the lumbar spine as a whole is approximately 55-60 degrees (Whiting & Zernicke, 1998).

2.2.2 Extension

Extension refers to the backward rotation of the trunk about the transverse axis. Motion of the spine in this direction is limited by the anterior longitudinal ligament, anterior aspects of the IV disc and the impingement of vertebral spines on each other (Kapandji, 1974). Extension at each joint in the lumbar spine has been reported to range from 1-24 degrees with greater values found at the more inferior aspects of the spine, while the extension limit of the lumbar spine as a whole is approximately 25-35 degrees (Whiting & Zernicke, 1998).

2.2.3 Lateral Bend

Lateral bending is described by a decrease in angle between the trunk and the pelvis in the frontal plane. Motion of the spine in this direction is limited by the supraspinous ligament, interspinous ligament, intertransverse ligament and the lateral aspects of the IV discs (Kapandji, 1974). Lateral flexion at each joint of the lumbar spine has been reported to range from 1-12 degrees, while lateral bending limit of the lumbar spine as a whole is approximately 20-30 degrees (Whiting & Zernicke, 1998).

2.2.4 Axial Rotation

Axial rotation can be described as twisting of the trunk about a longitudinal axis. Motion of the spine in this direction is limited by the supraspinous ligament, interspinous ligament, intertransverse ligament, torsion in the IV discs as well as the orientation of the facet joints (Kapandji, 1974). Axial rotation at each joint in the lumbar spine has been reported to range from 1-3 degrees while the axial rotation limit of the lumbar spine as a whole is approximately 5-12 degrees (Whiting & Zernicke, 1998).

2.3 LUMBAR SPINE KINETICS

When an external load is placed on the tissues of the lumbar spine, equal and opposite internal reaction forces must be developed by the muscles and passive tissues of the vertebral column if spinal stability is to be maintained. Further, if generation of movement about a joint is to be accomplished, internal forces such as muscle and joint shear, must exceed the force created by the sum of the external loads, the weight of the body, as well as gravity. These forces can be quite large depending on the loading characteristics and may have implications for injury of the various structures of the vertebral column.

2.3.1 Tissue Failure

When a load is applied to one of the body's tissues it will act to cause a change in that tissues size or shape. These changes in size or shape are known as strains and the internal reaction forces generated to counteract these strains are known as stresses. The degree of deformation of tissue following loading is dependent on the characteristics of the load,

such as magnitude and direction, and the properties of the tissue itself. Once the tissue is loaded beyond its elastic limit it will lose its ability to return to its original form once the load is removed. If the loading continues the tissue will reach its failure point, whereby it ultimately ruptures or tears (Watkins, 1999). Consequently, injury to the body's tissues occurs when its maximum strength, in terms of the stress/strain relationship discussed above, is exceeded. Further injury can occur in response to a single loading condition of high magnitude, but is more often the result of repeated conditions at low loads causing microtrauma of the body's tissues, thereby reducing its ability to resist further strains (Whiting & Zernicke, 1998). There are several mechanisms including compression and shear loading that can result in strain and thus injury of the body's tissues.

2.3.2 Moment

The moment of force can be defined as the effect of a force that acts to cause rotation about an axis. It is dependant on the magnitude of the force and the perpendicular distance from the axis of rotation. With respect to the joints of the body, moments created by the skeletal muscle act to control joint motion (Whiting & Zernicke, 1998). The weight of the body's segments, as well as external loads create moments at the joints, which must be countered by the tissues via a reactive or restorative moment. Since the moment arm of the muscles are much smaller than that of the external load, small external forces require large muscle and ligament in order to maintain postural stability (Timm & Malone, 1989). Further, as the external loads increase, skeletal muscle forces required to counter these loads increases as well (Watkins, 1999). These large forces required to maintain postural stability may come at the expense of an increased risk of

injury to the tissues of the lumbar spine. Norman et al. (1998), reported a significantly higher integrated lumbar moment (over an entire shift) for cases than controls, while Marras et al. (1993), found load moment to be related to an increased probability of high risk group membership for low back pain.

2.3.3 Compression

Spine compression is the reaction force, which acts to oppose the force that acts along the long axis of the spine, thereby making the intervertebral discs shorter and thicker. Under compression loading the first structure to fail is the vertebral body. Typically, damage is found at the endplate and is assumed to be caused by the annulus fibrosus of the adjacent disc bulging into the vertebrae (Adams & Dolan, 1995). That is, when the spine is loaded compressively, the nucleus accepts 1.5 times the externally applied load whereas the annulus accepts only half. Further, as the nucleus is unable to expand, loads are transmitted to the annulus in the form of tension, which are then transmitted, by the annulus to the superior and inferior vertebral endplates it merges with. Consequently, the tensile stresses radiating in equal partitions to all direction of the annulus are 4-5 times the externally applied load (Nachemson, 1975).

According to Moore and Garg (1992) failure of the lumbar spinal motion segments varies between compressive loads of 3 to 12 kN. Jäger and Luttman (1989) found the mean value at failure to be 4.4 kN, while Brinckmann et al. (1988) found ultimate compressive strength to average between 2.1 to 9.6 kN. Failure occurs at much lower loads when the spine has been loaded repetitively (Brinckmann et al., 1988; Adams & Dolan, 1984). Spinal compression forces have also been shown to increase with an

increase in the rate of load application, the weight of the load as well as horizontal distance of the load from the trunk (LeVeau, 1991).

Compression of the spine is the result of the combination of the weight of the upper body, acceleration of the upper body, the external load, and the tension generated by the muscles in an effort to maintain posture. However, according to Adams and Dolan (1995) most of the compressive forces acting on the spine are generated by the muscles. As a result, loads that require greater muscle activity will act to increase spinal compression and therefore the risk of injury to the vertebrae. This was confirmed by Callaghan and McGill (1995) in a lifting study where they found muscle activation levels to be greater for compression than shear loading trials. This increased level of activation in compressive trials was shown to result in a greater joint compressive penalty (Callaghan and McGill, 1995). Compression loading over an entire manual materials handling shift has been reported to be significantly higher amongst workers with low back pain compared to those without, and is therefore related to an increased risk of injury (Kumar, 1990; Norman et al., 1998). Norman et al. (1998) reported a mean cumulative compression estimate of 21 MN·s for workers with back pain and 19.5 MN·s for workers without back pain. Kumar (1990), reported a mean thoracolumbar cumulative compression estimate of 4.3 MN·s for male institutional aides without back pain and 11.1 MN·s for male institutional aides with back pain. Further, female institutional aides without back pain displayed thoracolumbar cumulative compression estimates of 6.0 MN·s and 7.9 MN·s for female institutional aides with back pain.

2.3.4 Shear

Shear can be defined as a load of two parallel forces, equal in magnitude and opposite in direction parallel forces that tend to displace one part of an object with respect to an adjacent part along a plane parallel to and between the lines of force (Watkins, 1999). With respect to the lumbar spine, shear force can be divided into three components. The load shear is the anterior shear force resulting from the load in the hands, the weight of the upper body as well as the acceleration of the load and the upper body. The load shear is resisted by the muscle shear and the joint shear forces. Muscle shear results from the restorative forces of the erector spinae. The joint reaction shear refers to the forces placed on the intervertebral joints as a result of the load shear that is not countered by the muscle shear (Potvin et al., 1991).

Normal lumbar lordosis naturally places the intervertebral disc and the pars under a shear load during standing. That is, as a result of the curvature of the spine, there is a tendency for the vertebral bodies of L4/L5 to slide forward and downward on L5/S1, thereby subjecting the L4/L5 and L5/S1 discs to considerable loading (Watkins, 1999). The main structure resisting the forward slipping motion is the pars. This is supported by Yingling and McGill (1999) who suggest that the main structure resisting shear is the intervertebral disc (70%), followed by the pars (30%), which is the structure most likely to fail during shear loading. The pars is able to withstand 2000 N of shear loading and deform between 8 mm and 10 mm before failure (Troup, 1976).

The magnitude of a load as well as the acceleration of the load and the body during manual materials handling tasks can result in shear forces approaching the failure tolerance limits of the facets. The low back musculature increases its force output in response to the load shear in an effort to keep joint shear within a margin of safety. As a result, peak shear forces that have to be supported by the intervertebral joints remain relatively constant at approximately 200 N, regardless of the load (Potvin et al., 1991). Joint shear loading of the lumbar spine over an entire shift, however, has been shown to be a risk factor for low back pain (Kumar, 1990; Norman et al., 1998).

The loading rate of applied shear forces has also been proven to affect the type of failure identified in the motion segment. That is, vertebral motion segments loaded slowly tend to exhibit fractures of the pars. However, vertebral motion segments loaded quickly tend to exhibit endplate avulsion fractures as well as fractures of the pars (Yingling & McGill, 1999).

2.3.5 Flexion

Trunk flexion can be explained as the decrease in angle between the trunk and the pelvis in the sagittal plane and involves both tension and compression of the structures of the intervertebral joint. Flexion of the trunk has been implicated as a risk factor for low back pain (Punnett et al., 1991). During flexion the apophyseal joints, the ligaments and the annulus act to limit ROM while the extensor muscles act to resist the forces of gravity and the external load thereby maintaining spinal stability (Timm & Malone, 1989). According to Adams and Dolan (1995) the intervertebral ligaments provide almost 70%

of the spines resistance to flexion with the remaining 30% coming from the disc. Therefore, under rapid flexion loading conditions the posterior spinal ligaments (interspinous and supraspinous) are the first to fail. Damage to these structures may compromise the stability of the spine and increase the risk of injury to other tissues, namely the intervertebral disc. Further, during prolonged loading, the ligaments may be lengthened due to the effects of viscoelastic creep, thereby increasing the laxity of intervertebral joints. Under normal circumstances, the ligaments act to minimize the amount of forward flexion allowed between adjacent vertebrae. However, when the ligaments are strained the range of motion increases and thus risk of injury to the disc does as well. When the vertebral column is flexed, the anterior portion of the disc is in compression and the posterior portion is in tension. This creates a low pressure area in the posterior region and a high pressure area in the anterior portion causing the disc to bulge. Over time, the posterior fibers of the annulus are weakened increasing the risk of tears and eventually prolapse (Adams & Hutton, 1984).

McGill and Norman (1986) suggested, however, that failure of the ligaments is not common and is more likely the result of a traumatic event than flexion loading. Consequently, there is somewhat of a disagreement within the literature with respect to flexion loading. Muscular components have been reported to generate approximately 99% of the restorative moment during bending trials, while the ligament contributed less than 1% (McGill and Norman, 1985). In fact, in their study the sacrospinalis muscle bore the greatest load of all the lumbar tissue and may therefore be at the greatest risk for injury during flexion. Further, in a study done by Potvin et al. (1991) the ligaments did

not contribute more than 60 Nm to the restorative flexion moment (recall that according to Adams and Hutton (1991) the first sign of damage to the ligament occurs in response to a passive bending moment of 60 Nm). However, although subjects were flexed at the trunk in both of these studies, the individual intervertebral joints were only moderately flexed due to the postures adopted by the subjects. That is, subjects were able to maintain a lordotic curvature of the spine, which would act to reduce the demand on the ligaments and increase the demands on the musculature. A middleground between the two extremes may be that the amount of flexion adopted by the lumbar spine will determine the contribution of the passive tissues. When the lumbar spine is in a neutral lordosis the passive tissues are not stressed and the muscle bears the responsibility of maintaining stability. However, there comes a point in the flexion arc where muscle activity ceases and spine stability becomes dependant upon passive structure and further loading will result in damage to the apophyseal joints (Timm & Malone, 1989).

2.3.6 Lateral Bend

Lateral bend can be defined as a reduction in the angle between the trunk and the pelvis in the lateral plane and has been implicated as a risk factor for developing low back pain (Punnett et al., 1991). EMG studies have reported that prolonged lateral bend loading conditions causing fatigue in the agonist, resulted in increased cocontraction of the antagonist muscles (Potvin & O'Brien, 1998). This increased muscle activation required to stabilize the spine, known as cocontraction, acts to increase spinal compression loads, which may increase the risk of injury. Andersson et al. (1974)

reported an increase in intervertebral disc pressure when the trunk was laterally bent 20 degrees.

2.3.7 Torsion

Axial twisting, or torsion, can be described as a rotation of the trunk about a long axis. Typically, a tissue subjected to torsion loading experiences tension, compression and shear (Whiting & Zernicke, 1998). Due to its structure, the intervertebral disc may be at risk for injury under this type of loading condition. In fact it is estimated that 90% of torque generated is resisted by the intervertebral disc, mainly the annulus (Farfan et al., 1972). The outer ring of the annulus is composed of several fibrous layers whose orientations oppose each other. As a result, when in torsion, half of the fibers of the disc are put in tension while the others become slack. Over time the fibers put in tension will begin to fail leading to herniation and prolapse of the disc. Farfan et al. (1972) have reported that torsion increases stress in posterolateral regions of the annulus, which is a common site of disc degeneration. In fact, twisting while lifting a heavy load more than 25 times per day results in a threefold increase in risk of acute prolapse of the lumbar intervertebral disc (Kelsey et al., 1984).

Other authors disagree with this mechanism of injury and suggest that the apophyseal joints are the main structure to resist torsion. The range of motion of the lumbar segments for torsion usually falls between 1 to 3 degrees (Adams & Hutton, 1981) and may not result in enough strain to damage the annulus. Instead, the apophyseal joints, which restrict the range of motion, may be at an increased risk for

injury as they are pressed together and apart on the contralateral side in this posture. These structures are able to resist shearing forces of about 2 kN (Cyron et al., 1976).

2.4 CUMULATIVE LOADING

Cumulative loading is a risk factor for low back pain (Kumar, 1990; Norman et al, 1999) and can be defined as the repeated application of a load or the constant application of a load for a prolonged period of time. This type of loading can result in accumulated microtrauma to the tissues of the body and leaves them less able to withstand subsequent loads (Whiting & Zernicke, 1998). Continuing to load these tissues will eventually lead to its failure. The number of loading cycles required is dependant upon the loading characteristics as well as the characteristics of the tissue itself.

2.4.1 Fatigue Failure

It has been reported that failure occurs at much lower loads during repetitive loading. Brinckmann et al. (1988) reported a 30% decrease in compressive strength of the vertebral body after 10 loading cycles. Adams and Dolan (1985) cyclically loaded lumbar motion segments with a compressive force oscillating between 500 N and 4000 N (representative of loading during normal activities such as standing and lifting). Of thirteen healthy motion segments, five failed at an average peak load of 2240 N. During testing of a second set of specimens, all 29 failed at an average peak load of 3800 N (Adams & Dolan, 1985). These values are well below the ultimate compressive strengths reported by Jäger and Luttman (1989) and Brinckmann et al. (1988) of 4.4 kN and 2.1-

9.6 kN, respectively. Further, both Hansson et al. (1987) and Brinckmann et al. (1988) found fatigue strength to decrease with an increase in magnitude of load.

2.4.2 Cumulative Loading and Low Back Pain

To date, there have been very few attempts at quantifying cumulative loading on the low back. Kumar (1990) investigated cumulative load amongst five institutional aides. Results indicated that cumulative compression and shear values were significantly higher in those aides who had reported pain in those who had not. In fact, male aides without pain experienced an average daily work exposure of 6.6 MN's of compression in the lumbosacral region of the vertebral column while aides with pain experienced an average of 15.4 MN's (Kumar, 1990). Norman et al. (1998) conducted a study where the cumulative loading experienced by workers in an auto assembly plant was estimated. Workers who had reported back pain prior to the study displayed significantly higher loading values on all biomechanical variables measured including lumbar moment, usual hand force, trunk velocity and compression and shear forces at L4/L5 (Norman et al., 1998). Lastly, Jäger et al. (2000) used a postural classification system to calculate the shift dose of compression, shear and moment on the lumbar spine in four occupations including field or surface construction, drop forge, industrial meat packing and refuse collection. These values were compared against the Mainz-Dortmund Dose Model, which recommends a critical dose of 5500 N/h over the course of a shift. This critical dose is based on a epidemiologic relationship between low back disease and cumulative lumbar load (Jäger et al., 2000). Incidentally, this threshold was exceeded in three of the four occupational fields studied.

2.4.3 Proposed Injury Mechanisms

Although there has been some consensus within the literature that cumulative or repetitive loading is a major risk factor for LBP, the mechanism through which it manifests itself remains unclear. Research within the body of literature seems to point to either the responses of the passive tissues of the spine or of the musculature. It has been demonstrated that cumulative loading can act to increase the ROM of the intervertebral joint. This laxity acts to reduce the stability of the spine and thus has implications for injury. In a study done by Goel et al. (1988) cyclic flexion of the ligamentous spine under small flexion moments resulted in a significant increase in ROM during extension activities, compared to preloading conditions. This finding suggests a partial loosening of the disc, since the disc, along with the anterior ligaments and facets are the primary structures resisting extension loads (Goel et al., 1988).

Solomonow et al. (1999) applied a cyclic passive load to the L4/L5 spinal motion segment of an anaesthetized cat. Reflexive muscular activity of the multifidus was recorded via EMG and exhibited a decline in muscular activity throughout the loading cycle. When the laxity of the passive tissues of the spine was offset by increasing the preload value, full restoration of EMG activity of the multifidus was exhibited. It was concluded, therefore, that creep in the passive tissue of the spine as a result of cumulative loading acts to desensitize the mechanoreceptors housed within, thereby reducing activity of the musculature. This decline in muscle activity results in spinal instability and thus a greater risk of injury (Solomonow et al., 1999).

Sparto and Panrianpour (1998) reported significant declines in erector spinae forces accompanied by significant increases of activity in the latissimus dorsi and external oblique muscles during repetitive extension activities. As the erector spinae musculature generates approximately ten times the force of the other musculature, it is believed that its decline simply offsets the increase in activity of the other muscles. Consequently, the results showed decreases in anteroposterior shear in half the subjects with varied compression results. Therefore, the hypothesis that an increase in muscle force, due to the coactivation of antagonist muscle groups results in increased spinal compression forces was not confirmed (Sparto & Parnianpour, 1998).

Neuromuscular fatigue has been defined as any reduction in the maximum force generating capacity of a muscle (Bigland-Ritchie et al., 1986), which can be demonstrated by a decrease in mean power frequency and increases in the amplitude of EMG signals (Petrofsky & Lind, 1978). Potvin and Norman (1993) reported a 21% decrease in erector spinae force during a 20 min lifting session, and 17% in a 2 hr session. This reduction in force may result in higher demands being placed on the passive structures of the spine. Other studies have reported that fatigue in the agonist musculature results in an increase in the activity of the antagonist musculature known as co-contraction. It is believed that co-contraction acts to stabilize the joint but comes at the cost of an increased joint compressive penalty, thereby increasing the risk of injury to the vertebral body endplates. Further, antagonist activity as low as 8% MVC has been shown to produce significant compression loads on the spine (McGill, 1992). Potvin and O'Brien (1998) measured co-contraction levels during fatiguing lateral bend exertions.

Average muscle activity when pooled across the musculature, increased by 17% for the agonists and 8% for the antagonists. In a similar study, prolonged isometric twisting resulted in an increased activation of antagonist trunk rotators thereby increasing the joint compressive penalty (Potvin & O'Brien, 1997).

The exact mechanism by which cumulative loading produces injury remains unclear and more research in the area is required. There may be more than one mechanism which is dependant on multiple loading characteristics. During repetitive flexion loading, spinal compression values are reported to increase (McGill, 1997; Granata & Marras, 1995) while during repetitive extension loading spinal compression values do not seem to consistently increase or decrease (Sparto & Parnianpour, 1998). Cholewick and McGill (1996) propose that stability is at its lowest and is maintained primarily by passive tissues during light activities while the opposite is true for more demanding activities. Consequently, the magnitude of the load applied over time may influence the contributions of various tissues and thus the mechanism of injury.

Lastly, the duration of the loading in combination with the other loading characteristics may influence the type and means of injury resulting from cumulative loading of the spine. Marras and Granata (1997) reported significant changes in lifting mechanics during extended periods of lifting cycles. Trunk ROM, velocity and acceleration decreased in the sagittal plane, while these same measures increased for the hip (Marras & Granata, 1997). Specifically, this change in mechanics resulted in a

decrease in spinal compression and increase in spinal shear forces, thereby placing different tissues within the same motion segment at risk for injury.

2.5 NON-OCCUPATIONAL ACTIVITIES

Historically, biomechanical analysis of the low back has only considered industrial activities. Despite the fact that biomechanical assessments of workplace tasks have improved recently, workers are still reporting low back pain. One major limitation of the work that has been done, is that only short duration evaluations of peak loads during work are usually collected. Spinal loading occurs throughout the entire day during work and non-occupational activities, but has had very little attention in the literature due to the high cost of data collection and analysis. Non-occupational activities refer to a wide range of activities such as childcare, housekeeping, gardening, and recreation require a wide range of trunk postures including flexion, lateral bending and axial twisting. Since we spend considerable time under load outside of work, the resulting loading mechanisms have implications for injury in the workplace, and thus, serious consideration is warranted.

With respect to childcare, a study was done by Grant et al. (1995) to assess the work activities and musculoskeletal complaints among preschool workers. They reported that 61% of respondents complained of back pain. A breakdown of activities was also performed and it was discovered that significant periods of time were spent kneeling, sitting on the floor, squatting and bending at the waist. These postures corresponded to activities such as feeding, changing and grooming children. Further, staff members who

work with smaller children performed more lifts than those working with older children putting themselves at greater risk for injury. Lifting and bending at the waist have already been identified as risk factors in previous sections, but sitting can also place high demands on the spinal segments (Hedman & Fernie, 1997).

In a study done by Hedman and Fernie (1997), L1-S1 vertebral columns were statically loaded in flexed and extended sitting postures. They found that the spinal segments experienced viscoelastic creep with greater displacement in the extended postures. As stated before, creep can increase the laxity of the intervertebral joint, thereby placing a greater demand on the spinal tissues. This is evidenced by a 183% increase in ligament tension for extension postures and 153% for flexion. Further, anterior column force increased 32% in flexed postures and 28% in extended postures over a thirty-minute period (Hedman & Fernie, 1997).

Housekeeping is another common facet of non-occupational activities. Several different types of activities fall into this category, although little work has been done in the area. Milburn and Barret (1999) reported that approximately 20 loaded forward flexion movements are required to make a bed. Further, L5/S1 moments produced during bedmaking may exceed safe lifting limits for certain height and size combinations (Milburn & Barret, 1999). In fact, peak compression forces measured during this experiment were found to exceed 6 kN.

2.6 EXPOSURE LIMITS

In order to reduce the incidence of LBP found in the workplace it is important to understand the types and magnitudes of loads that the tissues of the low back can sustain. Biomechanical analyses have been performed to test the strength of tissues of the spine during different loading conditions, and these results have been used to help form guidelines for safe working conditions during manual materials handling tasks.

2.6.1 Peak Loading

Typically biomechanical research into the sources of LBP has focused on peak loading conditions. Jäger and Luttman (1989) found the ultimate compressive strength of the lumbar spine to have a mean value of 4.4 kN while Brinckmann et al. (1988) found ultimate compressive strength to oscillate between 2.1-9.6 kN. An epidemiological study performed by Herrin et al. (1986) found that the rate of LBP reported by workers with jobs requiring compressive forces between 4.5-6.8 kN was 1.5 times greater than those with jobs requiring compressive forces less than 4.5 kN (Herrin et al., 1986). Further, Bringham and Gargt (1983) reported that jobs performed by workers with muscle strains had an average estimated compression force of 5.34 kN, while jobs performed by workers with disc injuries had an average estimated compression force of 7.97 kN.

2.6.2 Compression Tolerance Limits

Studies such as those discussed above have been used to establish compression tolerance limits for safe working conditions in the workplace. The National Institute for Occupational Safety and Health (1981), has proposed an Action Limit (AL) of 3400 N

and a Maximum Permissible Limit (MPL) of 6400 N. Jäger and Luttman (1991) propose a limit of 5700 N (+/- 2600) N for the average male and 3900 N (+/- 1500) N for the average female, while Mital et al. (1993) propose a limit of approximately 3900 N for males and approximately 2700 N for females.

Although there are a few proposed limits for some peak variables, ultimately, exposure limits for cumulative variables must be addressed. Peak loads at failure for spinal motion segments during repetitive compressive loading have been reported to occur at loads as low as 2240 N (Adams & Dolan, 1985). This value falls below the values for all the compression tolerance limits discussed above and thus the development of cumulative exposure limits is justified. These exposure limits should be used in conjunction with peak limits in an effort to reduce the incidence of LBP and injury. Certain tasks must be completed, however, before an exposure limit can be determined including: 1) assessment of possible methods used to quantify cumulative loads and their effect on cumulative exposure values must be performed (Callaghan et al., 2001) 2) loads encountered by the low back during the course of the day including work and non-occupational activities must be quantified and 3) a link between the level of exposure and an increase in low back pain and injury must be established.

2.6.3 Methods for Quantifying Cumulative Loads

Although research in this areas is limited, a number of methods of estimating the cumulative load on the lumbar spine have been developed. Kumar (1990) administered questionnaires whereby institutional aides reported all postures required to perform the

tasks associated with their jobs. These postures were input to a biomechanical model in an effort to obtain corresponding compression and shear force values. A cumulative loading value for each task was estimated by summing loading estimates for the posture at the beginning and end of each task as well as every 0.2 s between. Cumulative daily doses of compression were estimated by multiplying the task load by the magnitude of the load and summing each of the tasks together. In the same way, cumulative weekly, monthly and yearly doses can be extrapolated.

In a study done by Norman et al. (1998) cumulative loading of the lumbar spine of auto assembly workers was estimated. A trained observer watched the regular work duties of the employees and identified all instances of substantial loading of the lumbar spine. At this point the posture was captured on video and the observer recorded the duration, direction and magnitude of the load. This data was input to a 2D quasi-dynamic biomechanical model in an effort to obtain loading estimates of the lumbar spine. Peak loads obtained by the biomechanical model were multiplied by the duration of the task and the tasks summed together to estimate a cumulative loading profile of the lumbar spine for an average shift.

In a study done by Jäger et al. (2000) cumulative loading of the lumbar spine in four occupational professions was estimated. All tasks were captured on video and analysed via a posture classification system, which provided a detailed description of the position of various joints of the body, as well as the direction and magnitude of the load. This information was input to a biomechanical model in an effort to estimate loads placed

on the spine and a force weighted classification procedure was used to obtain shift doses of compression, shear and moment.

Electromyography (EMG) data can provide valuable information such as muscle activity as well as the relationship between muscle activity and force generation. Further, extrapolations can be made to estimate biomechanical variables from EMG data such as spinal compression. Potvin et al. (1990) presented a method of continuously estimating spinal compression values from EMG data (CNEMG) for assesement of jobs in the workplace. Subjects were asked to statically hold a 20 kg load with their torso slightly stooped for five seconds while EMG and video were recorded simultaneously. The compression mean generated by the biomechanical model was divided by the mean of the EMG for this task and multiplied by 100 in an effort to obtain a scale factor (Potvin et al., 1990). By multiplying the scale factor by the EMG data for each frame, compression for each frame, and thus an entire cycle, can be estimated. In this study, amplitude probability distribution functions were generated from the compression values, which made it possible to determine for which portion of the cycle the amplitude was above or below the various threshold limits such as the NIOSH AL and MPL (Potvin et al., 1990).

2.6.4 Comparison of Methods

All the studies discussed in the last few sections have employed different methods of quantifying a daily cumulative loading profile of the low back. This makes comparing values between studies challenging. It is necessary, therefore, to identify a common measurement technique that is practical to implement in the field and allows for minimal

error in the estimation of cumulative loading profiles. In a recent study done by Callaghan et al. (2001) the error present in five video-based approaches for estimating cumulative spinal loading was assessed. Specifically, five approaches were compared against the gold standard of rectangular integration of all frames collected at 30 Hz including: 1) rectangular integration at a reduced sample rate of 5 Hz 2) spinal loading at the initiation of the lift multiplied by the duration of the lift 3) cycle divided into work and rest 4) work phase only 5) cycle divided into four components including getting the load, lifting the load, placing the load and returning the load (Callaghan et al., 2001). Statistical analysis revealed that reducing the number of frame to 5 Hz was the only approach that did not result in significant error when compared against the gold standard of digitizing video at 30 Hz.

Although video analysis may be an accurate means of estimating cumulative loads and has the advantage of tracking shear and moment as well as compression, it may not always be a practical choice due to the substantial time and effort required in data processing, even at 5 Hz. EMG methods to estimate cumulative spinal compression loads have also been examined for certain tasks (Potvin et al., 1990; Mientjes et al., 1999). To date video based versus EMG methods for estimating prolonged loading cycles encompassing a wide variety of tasks, including non-occupational activities have yet to be compared.

CHAPTER III

3.0 METHODOLOGY

3.0.1 Video Method (Laboratory and Field Sessions)

All tasks in this thesis (field and laboratory studies) were recorded using a video camera in the sagittal plane. A 3D-image reconstruction system (Ariel Performance Analysis System) was used to obtain the position of the L4/L5 joint, hand, wrist, elbow, shoulder, ear and C7/T1 markers on the right side of the body. Tasks recorded in the field were manually digitised at 15 Hz (Callaghan et al., 2001), whereas those recorded in the laboratory were automatically digitized at 30 Hz. Joint coordinates as well as static hand loads were then input to a 2D static biomechanical model in order to estimate peak and cumulative L4/L5 moments, compression, joint shear and reaction shear forces. Peak estimates were identified as the highest loading estimate for each task. Cumulative loading estimates were calculated by the biomechanical model through rectangular integration of the force time history for each task.

3.0.2 Compression Normalised EMG (CNEMG) Method

Electromyography (EMG) data were collected using a ME3000 portable EMG system. Disposable surface electrodes (AgAgCl bipolar electrodes, Medi-Trace disposable ECG electrodes, Graphic Controls) were placed bilaterally over the thoracic erector spinae (TES). TES electrodes were placed 4 cm lateral to the spinous process at

the level of T9 (McGill, 1991). Ground electrodes will be placed over the bony portions of the spine in these areas. The portable EMG system sampled data at a frequency of 1000 Hz, rectified the sampled data and calculated the average value at a chosen averaging time of 0.1 s. An averaging time of 0.1 s takes the average of the raw values for 100 samples. This data was then manually low pass filtered at a cutoff frequency of 3 Hz using a second order butterworth filter. Only data from the TES muscles was collected as these muscles show less error with respect to the erector spinae when calculating the lumbar compression from EMG. This is probably due to an increased role of the extensor ligaments during flexion of the spine in the lumbar area relative to the thoracic area (Potvin et al., 1990).

3.0.3 EMG Normalization (Laboratory and Field Session)

In order to normalize the EMG data to compression loading estimates, measures of submaximal contraction were obtained. Subjects were asked to statically hold a 22 kg load with their torso inclined at approximately 60 degrees to the vertical for a duration of 5 seconds (Mientjes et al., 1999). EMG and video data were recorded simultaneously. These data were used to calculate a scale factor for converting lumbar erector spinae EMG from subsequent tasks into an estimated compression. This was accomplished according to the method outlined by Potvin et al. (1990), which required the following steps to be performed:

- 1) 500 N was subtracted from the actual L4/L5 compression force which was estimated from video techniques.
- 2) Average values of the new compression force (N) and EMG (mV) were computed over the duration of each lift.

- 3) A scale factor (N / mV) was calculated by dividing the compression mean by the EMG mean and then multiplying this value by 100.

3.1 DATA ACQUISITION

Data acquisition for this experiment was separated into two separate collections including: 1) a field collection session and 2) a laboratory mock-up session.

3.1.1 Field Collection

The field portion of this study was designed to quantify the cumulative moment, reaction shear, joint shear and compression forces on the lumbar spine over a two-hour duration and to compare video-based against EMG methods of estimating cumulative lumbar compression (see description for lab component section 3.1.2). Each subject was required to attend two sessions including an orientation/collection session as well as a mock-up session. The purpose of the orientation portion was to communicate the purposes and procedures to be followed during the study. At this time all questions were addressed and the consent forms were signed. The collection portion of the session was approximately two hours in duration and took place immediately after the orientation session within the subjects home. At this time subjects were fitted with surface EMG electrodes placed bilaterally over the TES, as well as the ME3000 portable EMG system (sections 3.0.1 and 3.0.2). Subjects were then required to perform the submaximal voluntary contraction trial (SMVC) for EMG normalization as outlined in section 3.0.3. This was required to calculate a scale factor for converting EMG data from subsequent tasks to estimates of spinal compression as per the method developed by Potvin et al

(1990). Once the SMVC trial was complete, subjects were asked to go about their normal daily routines. Subjects were followed and all tasks were recorded using a hand-held video camera. The purpose of this was to create a video journal of all tasks performed by the subject to be used for time keeping. Further, an assistant recorded the weights and dimensions of all loads handled by the subject during the collection session.

The video journal captured during the collection session was analyzed and twenty-one separate categories of tasks were identified (Table 1).

Table 1. Definition of all tasks identified from video journals for all subjects during two-hour collection session in the field study.

TASK	DESCRIPTION
Reaching floor to waist	All tasks performed between floor level and 89 cm. This category is comprised of several types of tasks such as retrieving household items from cupboards and changing the garbage bag.
Reaching waist to shoulder	All tasks performed at a vertical height ranging from 89 cm to 140 cm were added to this category. This category is comprised of several types of tasks such as washing dishes and wiping down counter tops.
Reaching above shoulder	All tasks performed at vertical heights above 140 cm were added to this category. This category is comprised of several types of tasks such as hanging clothes in closets and putting dishes away into cupboards.
Standing to sitting or sitting to standing	This category is composed of two similar tasks the first of which is the transition from a standing posture to the seated posture. The amount of time lapsed between the point where the subjects body leaves an upright standing posture to the point where the subject reaches a motionless seated position was recorded and added to this category. Similarly, the amount of time lapsed between the point where the subject leaves the seated position to the point where they reach an upright posture was also added to this category.
Sitting	This category is composed of the amount of time the subject spent in a seated posture
Standing	Standing was defined as an upright posture with both feet on the ground in the absence of any external load

Holding	Holding was defined as an upright posture with both feet on the ground in the presence of a load held close to the body.
Walking	Walking has been defined as an upright posture, where both feet are not in contact with the ground, in the absence of any load.
Carrying	Carrying has been defined as an upright posture, where both feet are not in contact with the ground in the presence of a load.
Vacuuming	Vacuuming has been defined as an upright or slightly stooped posture required to operate the vacuum machine. The use of vacuum attachments on stairways was not included in this category. Instead it was added to the floor to waist or waist to shoulder category depending on the dimensions of the task in question.
Crouching	Crouching is defined as a posture where both feet are in contact with the ground and the subject is flexed at the knee so that the back of the thighs are in contact with the back of the calves.
Standing to crouching or crouching to standing	This category is comprised of two postures the first of which is the transition from the standing to the crouching posture. This was measured as the amount of time lapsed between the point where the subjects body leaves a neutral upright posture to the point where the subject reaches the crouched posture. Similarly, the transition from crouching to standing was measured as the amount of time lapsed between the point where the subjects' body leaves the crouched posture to the point where the subjects' body reaches an upright standing posture.
Kneeling	Kneeling is defined as a posture where the feet are not in contact with the ground. Instead, the subjects' knees and shins are in contact with the ground.
Standing to kneeling and kneeling to standing	This category is comprised of two postures the first of which is the transition from standing to kneeling postures. This was measured as the amount of time lapsed between the point where the subjects body leaves a neutral upright posture and the point where the subject reaches the kneeling posture. Similarly, kneeling to standing was measured as the amount of time lapsed between the point where the subject leaves the kneeling posture to the point where they reach an upright standing posture.
Lifting With Two Hands	Two hand lifting was defined is the transition from the point w here the load leaves its initial start point to its endpoint while traveling in an upward posture while handling the load with both hands.
Lowering With Two Hands	Two-hand lowering is defined as the transition between the point where the load leaves its initial start point to its end

	point while traveling in a downward direction while handling the load with both hands.
Lifting With One Hand	One hand lifting is defined as the transition between the point where the load leaves the its initial start point to the point where it reaches it end point while traveling in an upward direction while handling the load with one hand.
Lowering With One Hand	One hand lowering is defined as the transition between the initial start point to its end point while traveling in a downward direction while handling the load with one hand.
Folding	Refers to an upright posture with both feet in contact with the ground. where the subject is folding an article of clothing.
Mopping	Refers to a slightly stooped posture required to push a mop back and forth.
Sweeping	Refers to an upright or slightly stooped posture required to manipulate a broom.

Once the video journal was analyzed and tasks identified the mock-up session was performed. Subjects were visited in their homes for a second time, where mock-ups of each task identified were performed. It was necessary to perform a mock-up session rather than digitize video captured during the collection session for three reasons:

- 1) A sagittal view of the right side of the body for video analysis was required. Due to spatial constraints within the home, this was not always possible during the collection session. Consequently, tasks were recreated as realistically as possible during the mock-up session in an effort to capture the required view.
- 2) During video analysis a stationary frame of reference is required. As the experimenter was following the subject with a handheld video camera, the frame of reference was constantly changing, and therefore not suitable for video analysis as the postures of the subjects would be skewed by the movement of the video

camera. The mock-up session allowed for the use of a tripod, thereby creating a stationary view of each task.

- 3) A fixed point within each frame of video was required to determine a scale factor during video analysis.

3.1.2 Laboratory Collection Session

The laboratory portion of the study was designed to quantify peak and cumulative loading estimates of low back moment, compression and shear as well as to validate the CNEMG method of estimating spinal compression against video techniques. Each subject was required to attend two sessions including an orientation session and a collection session. The purpose of the orientation session was to communicate the purposes and procedures of the study as well as to familiarize them with the tasks to be performed during the collection session. At this time, all questions were addressed and the consent forms were signed.

During the collection session subjects were fitted with surface EMG electrodes (AgAgCl bipolar electrodes, Medi-Trace disposable ECG electrodes, Graphic Controls) as well as an ME3000 portable EMG system as well as reflective joint markers at the hand, wrist, elbow, shoulder, ear canal, C7/T1 and L4/L5 on the right side of the body (Sections 3.0.1 and 3.0.2). Subjects were then asked to move into the testing area which was equipped with a fixed reference point required to provide a scale factor during video analysis as well as a lighting system. The lighting system was used to indicate the onset of EMG data collection and thus to synchronize video and EMG data in time. A

stationary video camera was left running for the duration of the session. This was used to capture a sagittal view of the right side of the subjects' bodies for each task. Subjects were required to perform two sets of the tasks outlined in Table 2 while EMG and video data were collected simultaneously:

Table 2. Description of tasks performed by subjects during laboratory collection session.

Submaximal Voluntary Contraction Trial	Subjects were asked to lift a 22 kg load from the ground, hold the load for approximately 5 s with their torso inclined approximately 60 degrees to the vertical, and then lower the load back to the ground. This was required to calculate a scale factor for converting EMG data from subsequent tasks to estimates of spinal compression as per the method developed by Potvin et al (1990).
Standing	Subjects were asked to stand with arms resting comfortably at the side for a duration of 5 s.
Folding	Two sweaters and a basket were placed on the ground in front of the subject inside the testing area. Subjects were asked to lift a sweater from the ground, fold the sweater, and then place neatly in the basket. This was repeated for the second sweater.
Basket	A laundry basket containing a 6 kg load was placed on the floor in front of the subject inside the testing area. Subjects were asked to lift the basket from the ground, hold for 3s, and then lower the basket back to the ground.
Garbage	A garbage can containing a bag with a 4.5 kg load was placed in front of the subject inside the testing area. Subjects were asked to remove the bag from the can and lower to the ground next to the can.
Iron	An ironing board was placed within the testing area and a shirt provided to iron. Subjects were required to iron the shirt for a duration of 10 s.

Shelves	Three loads, 4.5 kg, 2.2 kg and 1 kg were placed in front of a countertop (89 cm) within the testing area. Subjects were required to perform a series of lifts whereby they were required to lift each load from the ground to the countertop
Sitting	Subjects were asked to sit quietly with arms resting comfortably at their sides for a duration of 5 s.

3.2 SUBJECTS

For the field portion of the study three females and two males with no history of low back pain were recruited. Subjects had an average age of 29.6 +/- 9.9 years, height of 180.34 +/- 11.9 cm and mass of 69.5 +/- 29.3 kg.

Ten female subjects with no history of low back pain were recruited for the laboratory portion of the study. Subjects had an average age of 25.8 +/- 6.6 years, height of 176 +/- 3.3 cm and mass of 63.8 +/- 10.7 kg.

3.3 DATA TREATMENT

3.3.1 Video Data

Video records of each task performed in the laboratory and the field study were trimmed so that the beginning of each trial was determined to be the instant the trunk left the neutral posture. Similarly, the end point of each task was determined to be the instant the trunk returned to neutral posture. For the purposes of this study a neutral posture was defined as an upright trunk posture with the arms resting comfortably at the sides.

Subjects could be in a sitting, standing, crouching or kneeling position. Body landmarks (hand, wrist, elbow, shoulder, ear canal, C7/T1 and L4/L5) were automatically digitized at a rate of 30 Hz for the laboratory study and manually digitized at a rate of 15 Hz for

the field study. Video records were then digitally filtered at a cutoff frequency of 6 Hz using APAS software. These data were input into the GOBER static two-dimensional biomechanic model (Callaghan, University of Guelph) in order to yield peak and cumulative estimates of L4/L5 moment, reaction shear, joint shear and compression for each task. Rectangular integration of the force-time histories were performed to obtain cumulative estimates for each task while peak values were determined to be the highest loading estimate for each task. With respect to the laboratory study, sitting, standing and ironing tasks subjects were analysed for a duration of five seconds. All other tasks were analysed from the beginning of the task until the subject completed the task and thus time was not controlled. For the field session, cumulative loads associated with tasks performed more than one time were obtained by multiplying a single task estimate by the number of times a task was performed. The total exposure for two-hours was obtained by summing all cumulative load estimates for each task.

3.3.2 Electromyography Data

EMG data were full wave rectified and then low pass filtered at a cutoff frequency of 3 Hz using a second order Butterworth filter. The EMG data were then converted into a compression estimate for each frame by multiplying it by the scale factor determined by the SMVC trial and adding a 500 N passive bias to the EMG compression value. The bias is reflective of the compression forces acting on the discs of the spine due to the weight of the upper body as well as the load in the hands. A generic bias of 500 N was chosen as it minimized the errors in the EMG based compression estimate for tasks where subjects were handling a load (Potvin et al., 1990).

3.3.3 Submaximal Voluntary Contraction Trial (SMVC) Data

The purpose of the SMVC trial was to provide a scale factor for converting EMG data into compression as per the method developed by Potvin et al (1990). Only the portion of the task where the subject was holding the load (approximately 5 s) was considered during data treatment. Consequently, the trial was trimmed so that the start and end points corresponded to the duration of time the subject spent holding the load. The trimmed video was then digitized and filtered using APAS software as per the methods outlined above. The 500 N bias was subtracted from the actual L4/L5 compression force. It was necessary to subtract a passive bias as some of the compression force at L4/L5 is the result of the upper body mass and the load at the hands (Potvin et al, 1990). The scale factor was then calculated as per the methods outlined in section 3.1.3.

3.4 DEPENDANT VARIABLES

3.4.1 Field Study

Dependent variables for this portion of the study were divided into two categories:

EMG: 1) cumulative compression

VIDEO: 1) cumulative compression 2) cumulative reaction shear 3) cumulative joint shear 4) cumulative moment

3.4.2 Laboratory Study

Dependent variables for this portion of the study were divided into two categories:

EMG: 1) cumulative compression

VIDEO: 1) peak compression 2) cumulative compression 3) peak reaction shear 4) cumulative reaction shear 5) peak joint shear 6) cumulative joint shear 7) peak moment 8) cumulative moment

As stated earlier, for both the laboratory and field sessions, peak estimates were identified as the highest loading estimate for each task while cumulative loading estimates were calculated by the biomechanical model through rectangular integration of the force time history for each task. Further, the EMG model was used to calculate cumulative estimates of lumbar compression only.

3.5 DATA ANALYSIS

A repeated measures ANOVA was completed to determine if there were any differences between EMG and video based estimates of cumulative compression for each task in the laboratory session as well as between daily loading profiles in the field session. The model used for the analysis of the measured dependent cumulative compression consisted of a 2x1 within subject design. There was a single factor (method) with two levels (CNEMG, video). The statistical analysis was performed using a 0.05 level of confidence. The Pearson product moment correlation coefficient was also completed to determine the degree of correlation between EMG and video based estimates of cumulative compression at a 0.05 level of confidence. Linear regression plots were also calculated to display a visual representation of the correlation between methods for each task performed in the laboratory study.

Relative and absolute error between the two techniques were also calculated.

Absolute error was measured as the difference between the gold standard (video) and the CNEMG technique. Relative error was measured as the difference between the gold standard and the CNEMG technique divided by the gold standard.

CHAPTER IV

4.0 RESULTS

4.0.1 Field Study

Subjects spent an average of 23% of the two-hour collection session handling loads between the floor and waist, 20% handling loads between the waist and shoulder, 3% handling loads above the shoulder, 16% sitting, 6% standing and 6% walking (Figure 1). Subjects spent less than 5% of the session performing all other tasks. Table 3 displays the range of percent times for all tasks identified for each subject.

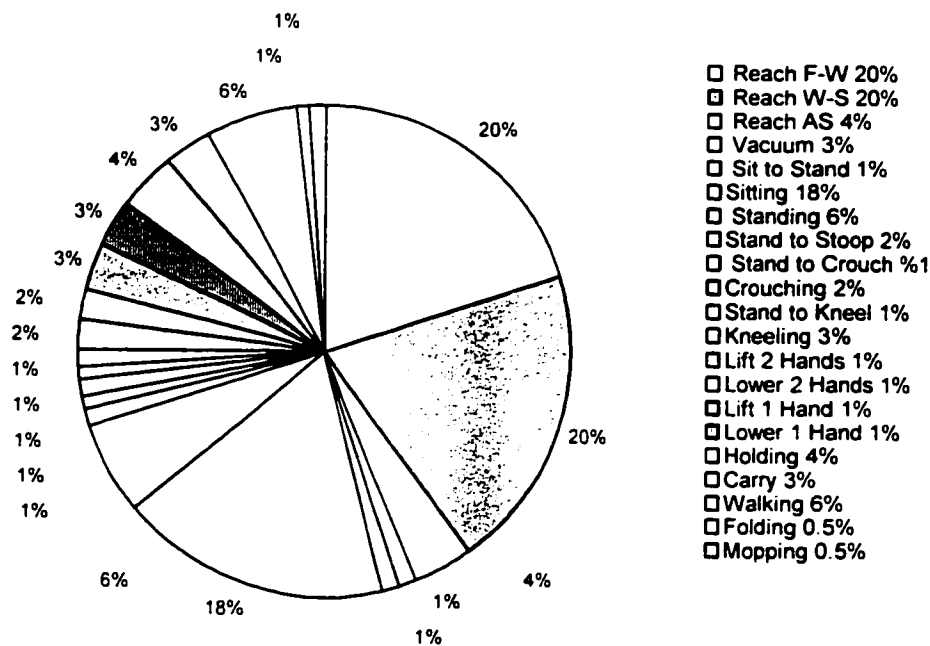


Fig 1. Task distribution (average time spent performing each task identified in video journals during two-hour collection session) for all subjects in the field study.

Table 3. Percentage of session spent performing each task identified in video journal of field collection for all subjects.

Task	Subject 1 % time	Subject 2 % time	Subject 3 % time	Subject 4 % time	Subject 5 % time	Average % time
Reach F-W	60.0	12.0	20.0	10.0	11.0	22.6 ± 21.3
Reach W-S	11.0	22.0	19.0	40.0	10.0	20.4 ± 12.1
Reach AS	1.0	6.0	1.0	4.0	3.0	3.0 ± 2.1
Vacuum	1.5	1.0	4.0	3.0	6.0	3.1 ± 2.0
Sit to Stand	0.5	0.5	1.0	0.5	0.5	0.6 ± 0.2
Sitting	5.0	14.0	19.0	16.0	26.0	16.0 ± 7.6
Standing	4.0	6.0	7.0	5.0	6.0	5.6 ± 1.1
Stand to Stoop	0.5	5.0	0.5	0.5	1.0	1.5 ± 2.0
Stooping	1.0	9.0	1.0	0.5	1.0	2.5 ± 3.6
Stand to Crouch	0.5	0.5	0.5	0.5	1.0	0.6 ± 0.2
Crouching	0.5	4.5	1.0	0.5	3.0	1.9 ± 1.8
Stand to Kneel	1.0	0.5	0.5	0.5	0.5	0.6 ± 0.2
Kneeling	0.5	6.0	1.0	1.0	8.0	3.3 ± 3.5
Lift 2 Hands	0.5	0.5	0.5	0.5	0.5	0.6 ± 0.2
Lower 2 Hands	0.0	0.5	0.5	1.0	0.5	0.6 ± 0.2
Lift 1 Hand	0.0	0.5	0.5	0.5	0.5	0.4 ± 0.2
Lower 1 Hand	0.0	0.5	0.5	0.5	0.5	0.4 ± 0.2
Holding	2.0	3.0	3.0	6.0	7.0	4.2 ± 2.2
Carry	4.5	3.0	6.0	3.0	6.0	4.5 ± 1.5
Walking	5.0	5.0	8.5	6.0	7.0	6.3 ± 1.5
Folding	0.0	0.0	4.5	0.0	0.0	0.9 ± 2.0
Mopping	0.0	0.0	0.5	0.5	1.0	0.4 ± 0.4

The purpose of this section was simply to identify the percentage of the session that each subject spent performing different tasks. As stated earlier floor to waist, waist to shoulder and above shoulder categories are composed of several different tasks. Each of these tasks exhibits a different loading profile, a factor which is addressed in the following sessions.

4.0.2 Cumulative Estimates of Compression, Moment and Shear Forces (Field Study)

Mean cumulative compression, reaction shear, joint shear and moment load estimates were 10.08 MN·s, 0.65 MN·s, 0.19 MN·s and 0.26 MN·s respectively (Table 4).

An example of the amplitude probability distribution function for cumulative lumbar compression can be found in the appendix.

Table 4. Cumulative estimates of compression, reaction shear, joint shear and moment for each subject in the field study using the video method.

Subject	Cumulative Moment (Nm·s)	Cumulative Reaction Shear (MN·s)	Cumulative Joint Shear (MN·s)	Cumulative Compression (MN·s)
1	0.23	1.50	0.19	20.63
2	0.31	0.65	0.14	8.10
3	0.22	0.57	0.19	5.78
4	0.30	0.47	0.25	9.77
5	0.25	0.28	0.16	6.10
Mean	0.26 ± 0.04	0.65 ± 0.47	0.19 ± 0.04	10.08 ± 6.1

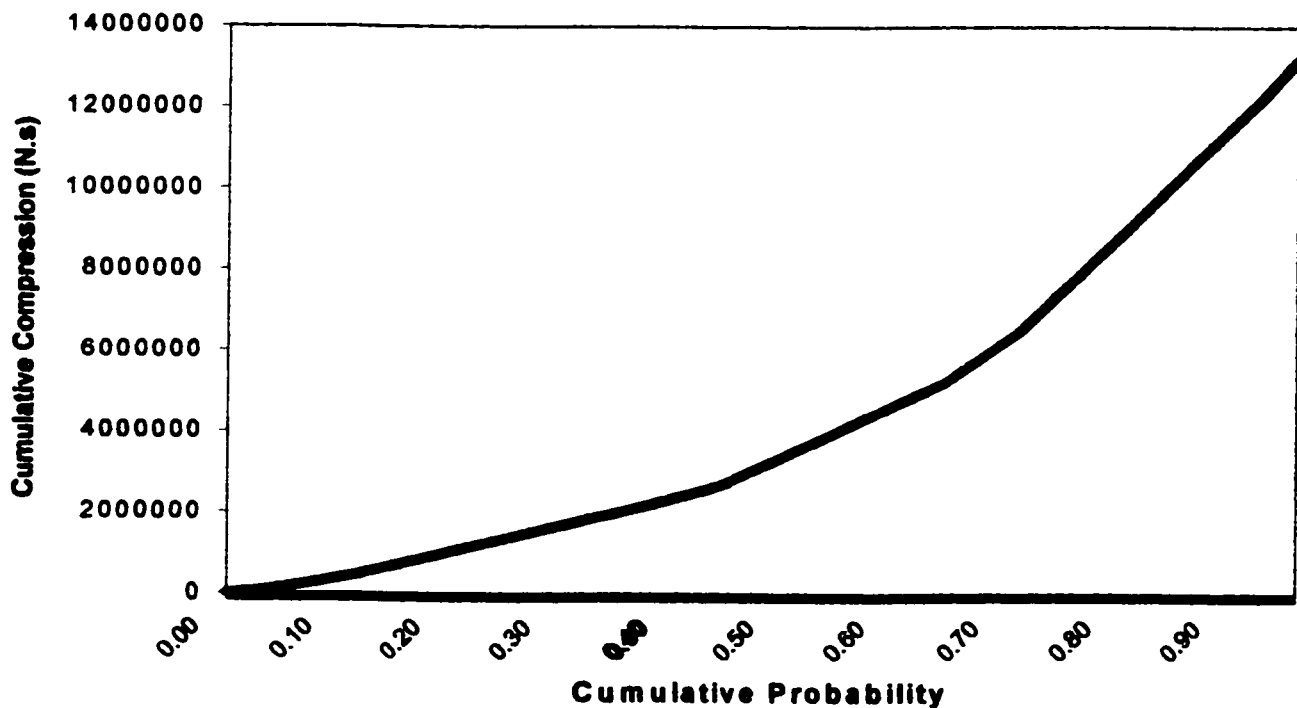


Fig 2. Amplitude probability distribution function for the field collection session using the CNEMG technique, subject 2.

4.0.3 Video Technique Vs CNEMG Methods of Estimating Cumulative Compression Forces (Field Study)

For the purpose of this study, estimates of cumulative compression obtained by video techniques were considered to be the gold standard as video techniques have been validated previously. Consequently, cumulative estimates of compression obtained by the CNEMG technique were measured against video techniques. The CNEMG technique exhibited a mean relative error of 33.7% when compared to video techniques for all subjects (Table 5). A repeated measures ANOVA revealed that significant differences did not exist amongst the mean values for the two techniques at a confidence level of $p < 0.05$. A correlation coefficient of $r = 0.96$ was determined which is also significant at a level of $p < 0.05$.

Table 5. Cumulative estimates of compression as determined by video and EMG methods during field study for all subjects. Absolute error was determined to be the actual difference between cumulative compression estimates for each method. Relative error was determined by subtracting the estimate determined by video methods from the estimate determined by EMG methods, dividing by the estimate determined by video methods and multiplying by 100.

Subject	Cumulative Compression Video (MN·s)	Cumulative Compression CNEMG (MN·s)	Absolute Error (MN·s)	Relative Error (%)
1	20.63	22.60	1.97	10
2	8.10	12.70	4.60	56
3	5.78	7.82	2.04	35
4	9.77	14.4	4.63	47
5	6.10	7.38	1.28	21
Mean \pm SD	10.08 \pm 6.11	12.98 \pm 6.17	2.94 \pm 1.58	34 \pm 18.70

4.1.1 Laboratory Study

Peak estimates of L4/L5 compression, moment, reaction shear and joint shear were determined using video methods only. Previous studies (Potvin et al., 1990) showed that the EMG method is not an accurate means of estimating lumbar spinal compression at a single instant in time. Based on this finding it is reasonable to assume that this method would not measure peak lumbar compression accurately. As a result EMG methods were used solely to calculate cumulative estimates of lumbar compression. Peak estimates of spinal compression estimated by video techniques ranged from 321 N observed during quiet standing to 3356 N during a lifting task in which a 4.5 kg load was lifted from the ground, held for 3 s and lowered back to the ground (Table 6). Peak estimates for shear determined by video methods ranged from 6.1 N to 427 N for the same tasks (Table 6). All subjects displayed peak compression estimates approaching (above 2000 N) or exceeding Mitals' tolerance limit of 2700 N for the basket task (Figure 2). Subjects 2, 4, 8 and 10 exhibited peak estimates approaching Mitals' tolerance limit for the folding task (Figure 2), while subjects 4, 6 and 8 approached the limit and subject 2 exceeded the limit for the garbage task (Figure 2).

Table 6. Mean (\pm SD) peak moment, reaction shear, joint shear and compression forces at L4/L5 for all tasks in the laboratory study using video techniques.

Task	Moment (N.m)	Reaction Shear (N)	Joint Shear (N)	Compression (N)
Stand	11 \pm 9	22 \pm 25	7 \pm 15	476 \pm 221
Sit	32 \pm 10	96 \pm 63	49 \pm 50	839 \pm 161
Fold	115 \pm 21	352 \pm 40	218 \pm 47	1979 \pm 358
Basket	141 \pm 21	408 \pm 28	212 \pm 39	2461 \pm 398
Garbage	117 \pm 31	347 \pm 45	178 \pm 38	2053 \pm 486
Iron	45 \pm 11	72 \pm 44	25 \pm 33	1078 \pm 190
Shelves	106 \pm 12	352 \pm 37	210 \pm 27	1930 \pm 193

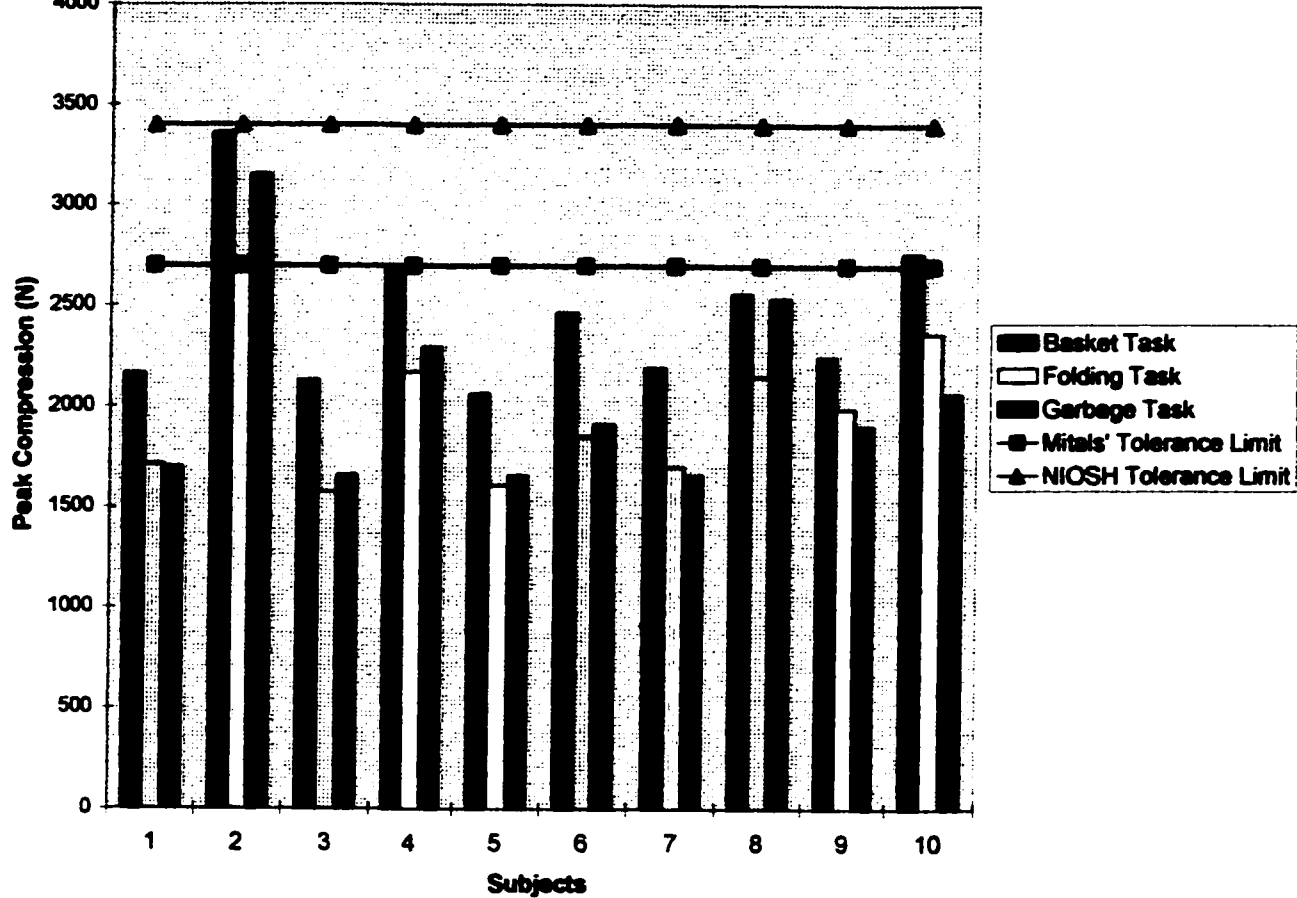


Figure 3. Average peak compression estimates determined by video techniques Vs Mital and NIOSH tolerance limits for the basket, folding and garbage tasks in the laboratory study (n=10).

4.1.2 Cumulative Estimates of Compression, Reaction Shear, Joint Shear and Moment Using Video Techniques (Laboratory Study)

Rectangular integration of the force-time histories was performed to obtain estimates of the cumulative loads for each task. Cumulative exposure compressive loads ranged from 2827 N·s during quiet standing for a 5s interval to 27783 N·s for a combination task of folding and low load lifting (Table 7). Shear exposure estimates for the same tasks ranged from 18 N·s to 3711 N·s (Table 7). As the amount of time required

to perform the task could not be controlled for all activities, cumulative exposures are a function of both load and time.

Table 7. Mean (\pm SD) cumulative moment, reaction shear, joint shear and compression forces at L4/L5 for all tasks in the laboratory study using video techniques.

Task	Moment (N·m·s)	Reaction Shear (N·s)	Joint Shear (N·s)	Compression (N·s)
Stand	32 \pm 28	83 \pm 55	53 \pm 41	2202 \pm 564
Sit	105 \pm 42	261 \pm 152	122 \pm 81	3370 \pm 714
Fold	775 \pm 223	2115 \pm 542	1192 \pm 246	17943 \pm 4243
Basket	560 \pm 94	1545 \pm 303	830 \pm 223	12288 \pm 1688
Garbage	660 \pm 298	1932 \pm 614	864 \pm 287	11998 \pm 4383
Iron	303 \pm 77	545 \pm 432	316 \pm 274	8203 \pm 1341
Shelves	853 \pm 164	2851 \pm 566	1577 \pm 314	17591 \pm 4560

4.1.3 Video Technique vs CNEMG Method of Estimating Cumulative Compression (Laboratory Study)

Cumulative estimates of L4/L5 compression from the CNEMG approach exhibited average relative errors of less than 10% for all tasks except sitting with the hands resting on the lap for a duration of 5 s and standing with the arms resting comfortably at the side for a duration of 5 s when compared to video methods (Table 8). The sitting task exhibited an average relative error of 57.6%, while the standing task exhibited an average relative error of 26.6%. When the 500N bias used to calculate the scale factor was adjusted to reflect the actual upper body weight of the subject, average relative error was reduced to 18.6% and 11.4%, respectively (Table 8). Figure 3 displays an example of the force time histories of the CNEMG and video techniques.

Table 8. Mean values and standard deviations for relative error between cumulative estimates via video analysis and CNEMG techniques for all tasks. All tasks but sitting and standing were calculated using a 500 N passive bias. The bias was adjusted to reflect the actual upper body weight of the subject for sitting and standing tasks.

Task	Video (N·s)	CNEMG (N·s)	Absolute Error (N·s)	Relative Error (%)
Basket	12288 ± 1688	13499 ± 1666	1321 ± 877	10 ± 5
Folding	17943 ± 4243	18904 ± 4282	1855 ± 678	10 ± 5
Garbage	11998 ± 4383	11593 ± 4133	1423 ± 1724	9 ± 8
Shelves	17590 ± 4560	17726 ± 4501	1673 ± 1015	10 ± 7
Iron	8203 ± 1342	8092 ± 1204	454 ± 318	5 ± 3
Stand	4159 ± 1115	4373 ± 900	430 ± 628	11 ± 18
Sitting	6444 ± 1618	7139 ± 1990	1097 ± 1706	27 ± 19

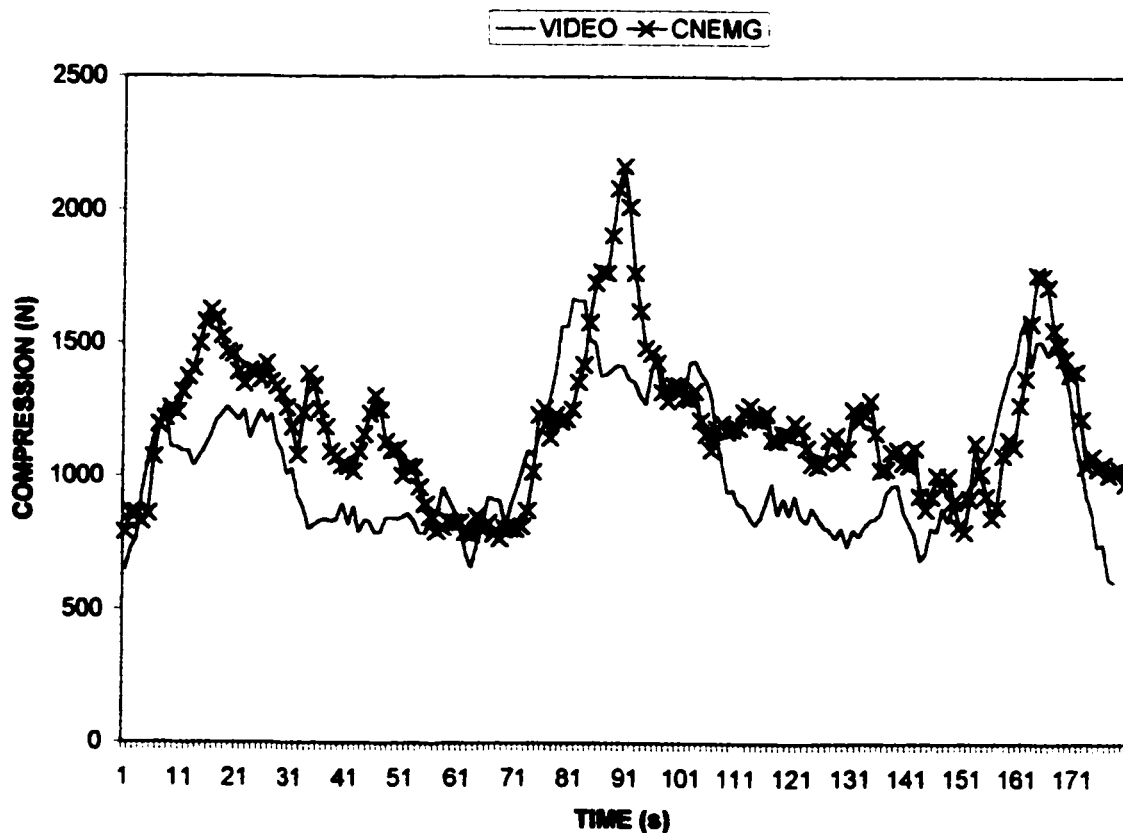


Fig 4. Force time history for CNEMG and video analysis techniques of estimating lumbar compression for the folding task, subject 1 (laboratory study).

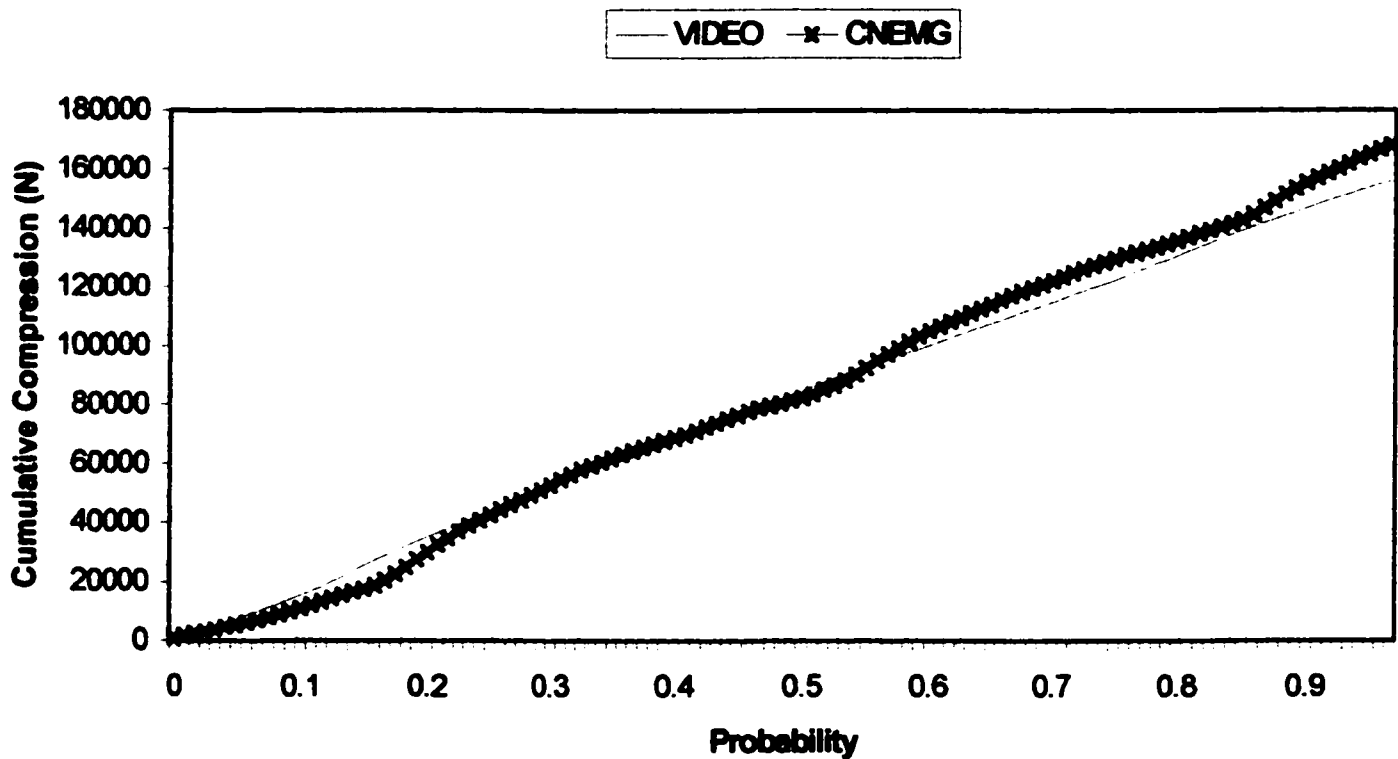


Fig 4. Amplitude probability distribution functions for video and CNEMG methods of estimating cumulative compression forces at L4/L5 for the shelves task in the laboratory study (subject 9).

Repeated measures ANOVA revealed that there were no significant differences between video and CNEMG techniques. All tasks were calculated at a confidence interval of $p < 0.05$. That is in all conditions but the basket task the null hypothesis stating that there is no relationship between CNEMG and video techniques was rejected at $p < 0.05$. Results of the Pearson product moment correlation coefficient analysis indicate that there is a significant correlation between the techniques used to estimate cumulative measures of compression about the lumbar spine for all tasks but sitting (Table 9). Further, when the correlation coefficient was calculated between methods for all of the tasks summed together a value of $r = 0.96$ was obtained which is also significant at $p = 0.05$.

Figure 5 displays a visual description of the positive correlation between methods for the basket task in the form of a scatterplot.

Table 9. Summary of significant findings for CNEMG and video techniques of estimating cumulative compression at L4/L5. * indicate that statistical techniques failed to show a significant correlation between EMG and video techniques.

Method	Basket	Fold	Garbage	Shelves	Iron	Stand	Sit
Pearson product moment correlation coefficient	$r=0.94$	$r=0.91$	$r=0.86$	$r=0.90$	$r=0.91$	$r=0.83$	$r=0.56^*$

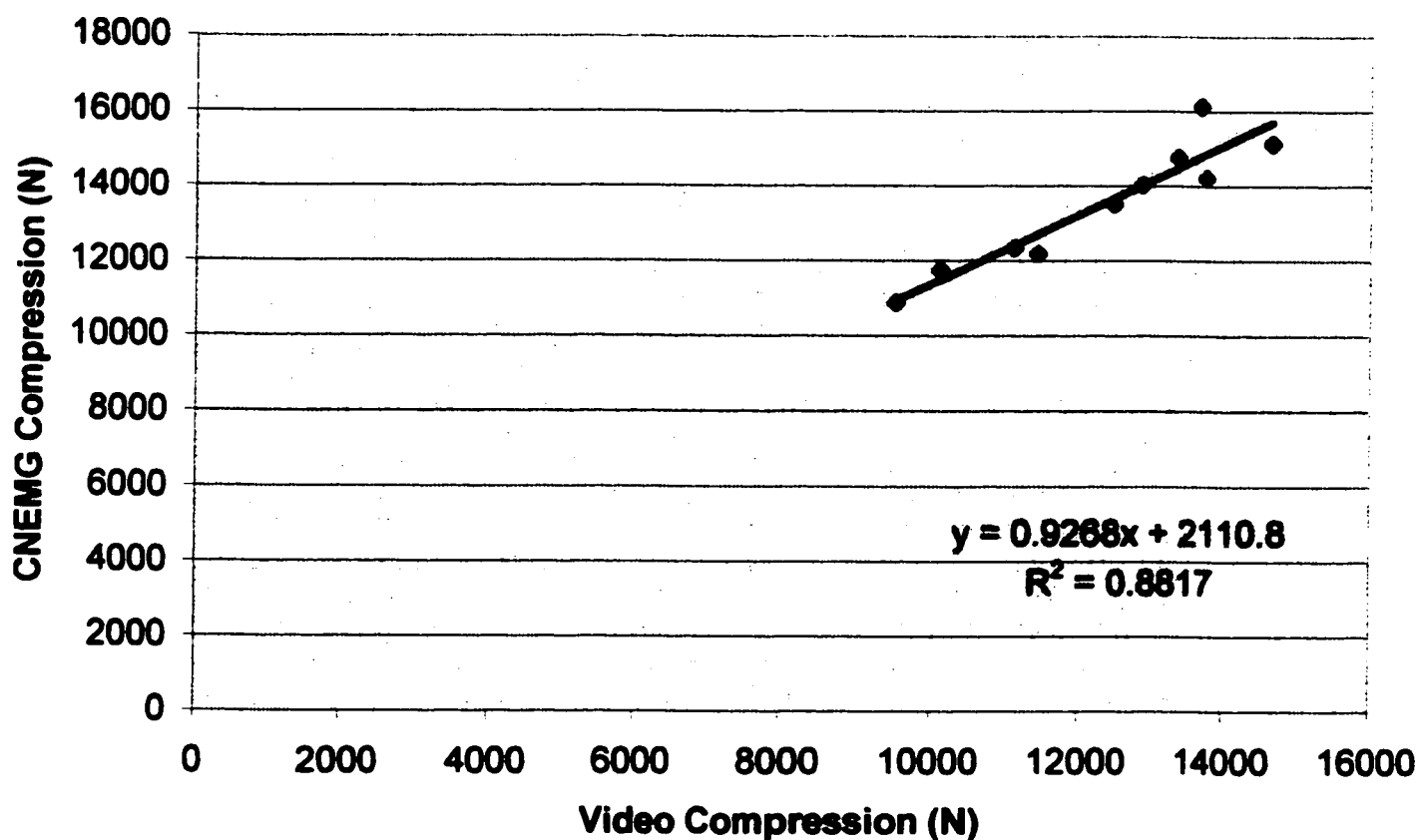


Fig 6. Linear regression plot for EMG and video methods of estimating cumulative lumbar compression. EMG and video values are average estimates for all subjects performing the basket task during the laboratory study (n=10).

CHAPTER V

5.1 DISCUSSION

5.1.1 Field Study

Currently there are no tolerance limits for cumulative compression, shear or moment exposure for the lumbar spine. Further, to the best of the authors' knowledge, cumulative low back loads that occur during daily non-occupational activities have not been presented previously. As a result, the degree of risk associated with loading estimates obtained in this study could not be quantified at this time. The majority of the loads handled in both the laboratory and the field collection session described in this study were below 10 kg suggesting that these tasks may be less physically challenging than many industrial tasks. This is somewhat misleading, in that considerable time outside of work is spent loading the spine in a variety of ways, and only a small sample of subjects and tasks have been analyzed thus far in the home. Further, low back spinal estimates in this study were obtained using a two-dimensional static model, which limited the number and types of tasks that could be incorporated. Further studies focusing on three-dimensional asymmetric tasks will undoubtedly result in greater loading estimates of the lumbar spine. This is because compression, moments and shear resulting from lateral bending or twisting cannot be calculated using a two-dimensional model and would therefore underestimate such values during asymmetric tasks.

Task distributions did show, however, that subjects spent an average of 23% of the collection session handling loads between the floor and the waist, 20% between the waist and the shoulder, 3% above the shoulder, 16% sitting, 5% standing and 5%

Walking. It is apparent that considerable time is spent at home involved in activities similar to those seen in all types of industry (e.g. sitting, standing and lifting). The duration for which subjects perform tasks between the floor and waist during non-occupational tasks, however, is a highly significant finding within itself. Performing tasks in this range requires trunk flexion, which acts to increase the moment about the lumbar spine. Consequently, the demands placed on the musculature and passive tissues at the joint required to resist the extensor moment are increased thereby increasing the joint compressive penalty at the lumbar spine. Within an industrial setting, specifically assembly operations, these postures associated with an increased risk of injury may not occur as frequently as non-occupational settings. Punnet et al. (1991) found that assembly workers in an automobile plant spend an average of 9.7% to 12.8% of the work cycle in mild flexion (21 to 45 degrees of forward flexion) and 4.1% to 7.4% of the work cycle in severe flexion (greater than 45 degrees of forward flexion). The difference between the amounts of time spent in a flexed posture may be the result of task layout within an industrial setting. That is although handling loads between the floor and the waist may be required for a small amount of tasks such as retrieving parts from bins, the majority of tasks are confined to the assembly line, which is typically located at approximately waist level. Within the home there are several tasks such as gardening, childcare, laundry and placing loads in shelving units, which require loads to be handled between the floor and the waist. Although the present study did not identify any tasks where subjects handled more than 10 kg, there are several instances where this could occur within the floor to waist range on a regular basis, for example lifting of small children and shoveling snow. In a study done by Kelsey et al. (1984) it was determined

that lifting loads greater than 11.3 kg more than 25 times per day was associated with a threefold increase in risk of injury to the intervertebral disc, especially if the knees were not bent during lifting tasks. Twisting of the trunk during lifting tasks, as frequently seen in non-occupational tasks acted to increase the risk of injury with less frequent repetitions (Kelsey et al., 1984). Other work settings, such as those in the health care industry may display similar frequency distributions with respect to handling loads between the floor and waist as those seen for non-occupational activities in this study. Just as in non-occupational tasks, workers are subject to a more random environment as the task characteristics are often determined by the height, weight and physical capabilities of the patients under their care. Further, these employees may be exposed to load handling activities in excess of compression tolerance limits proposed for use in industry on a regular basis, thereby significantly increasing their risk of injury.

Subjects exhibited a mean cumulative compression estimate at L4/L5 during non-occupational activities of 10.08 MN·s for a two-hour duration. This is equal to almost half of the daily estimates (8 hours) of L4/L5 compression for automobile assembly workers with low back pain as determined by Norman et al. (1998). Further, cumulative estimates of moment, reaction shear and joint shear determined in this study during a two-hour collection session were also equal to almost half the loading estimates determined by Norman et al. (1998) for the same variables. If values calculated in the present study were extrapolated to represent an eight-hour workday, they would be equal to almost double those observed in the Norman study (1998). The differences between techniques used to calculate a daily dose of lumbar compression should be noted.

Norman (1998) calculated loading estimates for working postures (instances of high spinal moments resulting from forward inclined trunk postures and/or high forces on the hands) as well as an upright standing posture to fill in time between working postures. As a result, any spinal loading that would fall between standing and working postures, which would inevitably act to increase the cumulative estimates, was lost. Further, any incidence of sitting which consistently produces higher estimates of lumbar compression was not accounted for in this study. The decision to exclude the medium levels of compression may have been made to reduce time spent collecting and processing data. Further, there is some evidence that some low levels of loading will never result in injury regardless of the length of exposure. Callaghan & McGill (2001) showed that spinal motion segments cyclically compressed at 260 N were still intact after 86, 400 cycles. In contrast, the present study accounted for every second of the collection session exactly the way it occurred. Consequently, the presence of loading conditions that fall between standing and working postures may account for the large difference in estimates.

Subject one exhibited a cumulative compression estimate of 20.63 MN·s, which is more than double the loading estimate for all other subjects. The increased compressive force about the lumbar spine can be explained by considering the task distributions. Subject one spent 60% of the collection session handling loads between the floor and waist while all other subjects spent less than 20% of the session handling loads in this range. As stated before, this stooped posture corresponds to an increased joint compressive penalty at the lumbar spine. Consequently, the amount of time spent handling loads in this posture is reflected by the cumulative compression loading

estimate. The range between cumulative compression estimates for subject one and all other subjects can be largely attributed to the amount of time this subject spent gardening. This finding also highlights the need for more research to be performed in this area. That is, studying a subject for a duration of two hours is not likely to reflect an average loading estimate if a subject was to be tracked for a prolonged period of time. One reason for this would be seasonal changes in the types of tasks people perform in their homes. As subject one illustrated summer and spring seasons are often associated with gardening and other yard work tasks while the winter is associated with other tasks such as shoveling snow and salting walkways. These tasks are performed in addition to any housekeeping or childcare activities that are performed consistently throughout the year, which will act to alter the loading profiles for the lumbar spine. Further, day-to-day requirements with respect to non-occupational activities vary within themselves. That is, there are some tasks that are required to be performed on a regular basis to maintain the home such as vacuuming, washing dishes or cutting the grass. Conversely, there are some tasks that are performed less frequently such as washing floors, shampooing carpets or moving furniture. As a result, different combinations in the types of tasks performed will alter the loading profiles for the lumbar spine. The behavior of the subject was also affected by the presence of an experimenter with video camera in hand. Although subjects were instructed to go about their normal daily routines and a period of acclimatization was provided prior to data collection it seems that subjects performed more work than they might normally perform over a two-hour duration if in private. A single study is, therefore, not enough to make global statements about cumulative loading in a non-occupational setting.

Cumulative exposure guidelines would identify the threshold level of loading at the lumbar spine that could occur before risk of injury increases. By ensuring that the amount of loading that occurs during work hours remains below this value is not enough. Loading of the spine occurs continuously throughout the day during work as well as in the home. This study showed that there is a considerable amount of spinal loading that occurs during non-occupational tasks, and should therefore be considered when implementing cumulative exposure guidelines. That is, if further research could identify an average estimate of spinal loading during non-occupational tasks, the amount of loading that should be allowed in industry over an eight-hour shift could then be adjusted so that the total amount of loading experienced by the spine is within recommended ranges for the entire day. There are also other variables that would act to alter the amount of cumulative exposure to loads at the lumbar spine during the workday. The effects of shiftwork may alter the ratio of loading during work activities to loading during non-occupational activities. That is, people working midnight shifts may have more time during the day to perform tasks within the home when compared against people on day or afternoon shifts. This would act to increase the cumulative spinal estimates observed during non-occupational tasks, thereby reducing the amount of cumulative loading that they can be exposed to during working hours. Further, the type of work setting these guidelines are to be used in should be considered. People working in an assembly plant may not handle extremely heavy loads on a regular basis. Punnett et al. (1991) performed a study where only 13% to 22% of subjects were observed handling loads 44.5 N or greater. In contrast, employees in a health care setting may be asked to

handle loads equal to or in excess of their own body weight on a regular basis, particularly during patient handling activities. Further, the differences between task distributions and thus loading estimates between males and females during non-occupational settings should be explored. In the present study, three females and two males were studied over a two-hour duration in a field setting. All female subjects exhibited higher cumulative compression, moment, reaction shear and joint shear estimates when compared against their male counterparts, largely due to an increased incidence of sitting and standing in the absence of a load.

➤

For the purpose of this study, cumulative estimates of L4/L5 compressive forces obtained by video analysis techniques were considered to be the gold standard. In order to validate the CNEMG method, cumulative L4/L5 compression estimates obtained by this method were compared against the gold standard. Relative error between the two techniques ranged from 10 % to 56 % between subjects. On average the CNEMG method had a tendency to overestimate cumulative lumbar compression estimates by 33 %. Consequently, the CNEMG technique may estimate lumbar compression at a single instant by approximately 400 N. Despite this, statistical analysis using repeated measures ANOVA indicated that there was not a significant difference between techniques. Although statistically, there is not a significant difference between techniques in absolute terms this method may not be an accurate method of estimating lumbar compression, as 400 N is approximately equal to the upper body weight of many of the subjects in this study.

As stated earlier the CNEMG method overestimated the cumulative lumbar compression by an average of 33 % when compared against the gold standard of video. The term gold standard implies that the video method is the most accurate method of estimating spinal loads. This is true only in the sense that the video method has been validated previously. A two-dimensional biomechanical model was used in this study to estimate cumulative estimates of lumbar compression. As stated earlier, this model cannot account for compressive forces resulting from lateral bending or twisting. As subjects were free to go about their normal daily routines, several asymmetric tasks were observed for each subject during collection sessions. As a result, the video method most likely underestimated compression estimates at the lumbar spine in these instances. Consequently, EMG methods may be a more accurate method of estimating cumulative compression for the tasks and subjects analyzed in this study. That is, EMG methods directly measured activity from the muscles during both symmetric and asymmetric activities for conversion to estimates of lumbar compression. Taking the average between sets of bilateral electrodes acts to minimize errors in estimates of cumulative lumbar compression. That is in the laboratory study it was observed that tasks where the load was distributed unequally between the hands electrodes on the side of the body loaded more heavily overestimated the lumbar compressive load. Further, loads on the side of the body handling the minimal amount of the load underestimated the lumbar compressive load. This unequal distribution of muscle activity on the right and left side of the body may be representative of the loading profile for an asymmetric task. Further, the video-based method in this study proved to be very restrictive for data collection in the home environment. Mock-ups were required to get the appropriate view

of all tasks performed, as many tasks occurred in tight physical spaces that did not allow for the appropriate views to be obtained by video. Although, care was taken to ensure mock-ups were performed as realistically as possible, inevitably some time was lost as well as error present with this method. In contrast, the EMG method collected data continuously while tasks were actually being performed by subjects. Although this method has not been validated for asymmetric tasks, based on these findings the CNEMG method of estimating cumulative lumbar compression may actually be the gold standard for this particular study.

5.2 Laboratory Study

The laboratory portion of the study was conducted to document peak and cumulative estimates of moment, reaction shear, joint shear and moment as well as to validate the CNEMG method of estimating L4/L5 compression against video analysis techniques. Two subjects exceeded Mitals' peak compression tolerance limit of 2700 N during a task where they were required to lift a 6 kg basket of clothes, hold for three seconds and then lower the basket to the ground. Further, four of the tasks, which required the subjects to handle no more than a 4.5 kg load displayed peak compression loading estimates approaching 2700 N. These peak compression estimates were similar to those reported by Adams & Dolan (1985) to result in failure of the lumbar spine during cyclic compressive loading. This is a significant finding as these types of low loading conditions are common to activities of daily living, and may have implications for the onset of low back pain and discomfort. Currently, there are no proposed tolerance limits for cumulative exposure and therefore comparisons could not be made for these values.

Comparison between video analysis techniques and CNEMG methods of estimating cumulative compression resulted in relative errors of 10% or less for all tasks but sitting and standing. The CNEMG technique had a tendency to overestimate lumbar compression for these tasks. When the 500 N bias used to calculate the scale factor was adjusted to represent the actual weight of the upper body relative error was reduced to 11% and 18.6% for the standing and sitting tasks respectively. It is important to note that almost half the error can be attributed to subject 7, who showed a relative error of 88% between techniques even after adjustments were made to the bias. A possible explanation for the large error estimate is an increased level of muscle activation in the thoracic erector spinae in order to stabilize the trunk while sitting. This increased muscle activity would not be realized by the video techniques and would therefore result in a large error estimate between the two techniques. When the error between techniques was calculated in the absence of subject 7, average relative error was reduced 11%. Results of the correlation analysis showed a statistically significant correlation between the two techniques for all tasks.

Previous research (Mientges et al., 1999) found that although the CNEMG technique was a viable method of estimating cumulative compression for symmetric tasks there was significant error present predicting cumulative loads for an asymmetric task. In this study, certain tasks required one side of the body to support the load for example ironing or lifting garbage with most of the weight supported by one hand. In these instances electrodes located on the side of the body supporting the greater amount of the load overestimated the compression profiles, while electrodes on the side of the body free of the load underestimated the compression profiles, a pattern which may also be

representative of asymmetric tasks. Calculating the average from the two sets of electrodes resulted in a strong correlation, $r=0.80$ or greater, for all tasks but sitting. As stated before when subject 7 was not included in statistical analysis the correlation coefficient was increased from $r=0.56$ to $r=0.82$ for the sitting task.

5.3 Hypothesis

1. Peak Measures of Non-Occupational Activities vs Occupational Activities

It was hypothesized that peak low back spinal loads estimated during non-occupational activities would be similar in magnitude to those reported for many work activities involving manual materials handling tasks. Norman et al. (1998) reported peak compressive estimates ranging from 2733 N to 3423 N for automotive assembly workers without back pain and with back pain respectively. Further, in a study done by Punnet et al. (1991) the jobs of 95 automobile assembly workers were analyzed for postural and lifting requirements using video techniques. Less than 3% of the analysed postures resulted in peak compressive forces of 3400 N or greater, the value equivalent to the action limit recommended by NIOSH (1981). The highest peak compressive forces estimated on these jobs was 5337 N (Punnet et al., 1991). Although none of the postures analysed in the present study exceeded the NIOSH action limit of 3400 N, 5% exceeded Mital's female tolerance limit of 2700 N, and 4% exhibited peak compression estimates approaching (greater than 2500 N) this critical value. According to the present study, when compared to industrial tasks, non-occupational activities may exhibit a greater incidence of postures associated with an increased risk of injury to the low back. This theory was not tested statistically, but instead is an observation based on the number of industrial tasks reported in the literature to exceed tolerance limits (3%) when compared

to the number of non-occupational tasks exceeding limits in the present study (5%). With respect to absolute magnitude of L4/L5 compression force, according to the literature, work activities appear to exhibit loading profiles greater than those of the non-occupational activities analysed in the present study. As stated before, this is somewhat misleading as only a small number of subjects and tasks have been studied thus far. Future studies incorporating a wider range of non-occupational tasks will undoubtedly result in greater loading estimates. Although, work and non-occupational activities could not be compared statistically due to a large difference in sample size the hypothesis cannot be accepted at this time.

2. Cumulative Measures of Non-Work Activities vs Work Activities

It was hypothesized that estimates of cumulative compression would be lower for non-occupational activities than for industrial activities. In a study done by Jäger et al. (2000), cumulative compression estimates ranged from 2.9 kN·s over a 3 hr period in a drop hammer task to 7.3 kN·s over a 5.6 hr period during a surface construction task. For this study, working situations were only analysed when the load in the hands weighed at least 0.5 kg or when the trunk was distinctly inclined, twisted or bent laterally. Kumar (1990) reported cumulative daily compression estimates in institutional aides ranging from 6.6 MN·s to 14.5 MN·s. Again, only tasks reported by the subjects were considered in this study. The present study reported cumulative compression estimates for a duration of 2 hrs while performing non-occupational activities ranging from 5.78 MN·s to 20.63 MN·s with a mean value of 10.08 MN·s, considerably greater than the estimates associated with industrial activities. Norman et al. (1998) reported cumulative

compression estimates ranging from 19.5 MN·s to 21.0 MN·s over an 8 hr duration for male institutional aides without and with low back pain respectively. The large variance may lie in the selection of tasks chosen to be included in the calculation of the daily loading profile. That is previous studies calculated only those working postures where the trunk was deviated significantly from neutral or there was a load in the hands, while a generic standing posture was used to fill in the rest of the time for their calculations of daily doses of compression. Consequently, all loading scenarios associated with compression estimates that fell between these two extremes were underestimated. Further, sitting which results in a greater compression estimates than standing was ignored. As the present study considered all postures as they occurred including those typically associated with rest (e.g. sitting) a greater daily dose of compression relative to those measured in industry was obtained. Further, the amount of rest associated with non-occupational activities was not all that different from manual materials handling tasks. On average subjects spent 16% of the session sitting, while the rest of the session was spent performing various tasks associated with housekeeping. The average industrial worker receives an hour of break time over an eight-hour shift (equal to two fifteen minute breaks and a thirty minute lunch), approximately 12.5% of the work-day. Many workers also have down time between work cycles, which would increase the percentage of rest over the duration of the day. Because the methods by which compression profiles were calculated differ, a true statistical comparison cannot be made between work and non-occupational activities. However, based on the values above the hypothesis is rejected.

3. CNEMG vs. Video Analysis Methods of Estimating Cumulative Compression Profiles

It was hypothesized that CNEMG methods and video analysis methods of estimating cumulative compression would not be significantly different from each other. The present study found that there was no significant difference between techniques for all conditions but the basket task at a confidence level of $p < 0.05$ as determined by repeated measures ANOVA. Pearson product correlation coefficient analysis indicated a statistically significant relationship between techniques for all tasks studied. The hypothesis was therefore accepted.

CHAPTER VI

6.1 CONCLUSIONS

Cumulative estimates of compression, joint shear, reaction shear and moment collected over two hours in the home were equal to almost half the cumulative estimates for the same variables collected over an eight hour shift in a previous study of industrial work. Subjects spent a substantial amount of time at home (20% of the 2 hour session) in postures associated with an increased joint compressive penalty at the lumbar spine (handling loads between the floor and waist). In the laboratory, a number of light lifting tasks were found to result in peak compression estimates with magnitudes similar to those that have been found previously to result in failure of the lumbar spine during repetitive compressive loading. From this work, it appears that further documentation and consideration of non-occupational tasks for the development of cumulative exposure guidelines for use in industry seems warranted.

On average, the CNEMG method substantially overestimated cumulative compression profiles determined from video records of non-occupational tasks in the home despite no significant statistical difference between the two methods. Conversely, the agreement between CNEMG and video methods for controlled laboratory tasks was quite good, suggesting that CNEMG may be an accurate approach for estimating cumulative L4/L5 compression during symmetric non-occupational activities when certain guidelines are followed. That is, when the load is distributed unevenly on one side of the body, the average between pairs of electrodes on the right and left side of the body should be calculated. Additionally, the bias used should reflect the actual upper body mass of each individual. Taken together, it is suggested that further research is

required to determine whether the CNEMG method can accurately predict cumulative compressive loads during field assessments of non-occupational tasks that involve significant asymmetry.

References

- Adams, M.A. and Dolan, P. (1995) Recent advances in lumbar spinal mechanics and their clinical significance. *Clinical Biomechanics* 10, 3-19.
- Adams, M.A. and Dolan, P. (1991) A technique for quantifying bending moments acting on the lumbar spine in vivo. *Journal of Biomechanics* 24, 117-126.
- Adams, M.A. and Hutton, W.C. (1985) Gradual disc prolapse. *Spine* 10, 525-531.
- Adams, M.A. and Hutton, W.C. (1981) The relevance of torsion to the mechanical derangement of the lumbar spine. *Spine* 6, 241-247.
- Adams, M.A. and Hutton, W.C. (1980) The effects of posture on the role of the apophyseal joint in the role of resisting intervertebral compressive forces. *Journal of Bone and Joint Surgery* 62-B, 358-362.
- Andersson, G. (1976) Low backache in working life. *Lakartidningen* 14, 1305-1307.
- Bigland-Ritchie, P., Cafarelli, E. and Vollestad, N.K. (1986) Fatigue of sub maximal static contractions. *Acta Orthopaedica Scandinavia Supplement* 128, 137-148.
- Brinckmann, P., Biggemann, M. and Hilweg, D. (1989) Prediction of compressive strength of the human lumbar vertebrae. *Spine* 14, 606-610.
- Brinckmann, P., Biggemann, M. and Hilweg, D. (1988) Fatigue fracture of the human lumbar vertebrae. *Clinical Biomechanics* 3, Supplementary 1.
- Bringham, C.J. and Garg, A. (1983) The role of biomechanical job evaluation in the reduction of overexertion injuries: a case study, paper at the 23rd Annual American Industrial Hygiene Association Conference, Philadelphia, PA.
- Callaghan, J.P. and McGill, S.M. (1995) Muscle activity and low back loads under external shear and compressive loading. *Spine* 20, 992-998.
- Callaghan, J.P. and McGill, S.M. (2001) Intervertebral disc herniation: studies on a porcine model exposed to highly repetitive flexion/extension motion with compressive force. *Clinical Biomechanics* 16, 28-37.
- Callaghan, J.P., Salewytch, A.J and Andrews, D.A. (2001) An evaluation of predictive methods for estimating cumulative spinal loading. *Ergonomics* 44, 825-837.
- Cholewicki, J. and McGill, S.M. (1996) Mechanical stability of the in vivo lumbar spine: implications for injury and chronic low back pain. *Clinical Biomechanics* 11, 1-15.

- Cyron, B.M., Hutton, W.C. and Troup, J.D.G. (1976) Spondylolytic fractures. *Journal of bone and Joint Surgery* 58-B, 462-466.
- Farfan, H.F. (1975) Muscular mechanics of the lumbar spine and position of power and efficiency. *Clinical Orthopaedics North America* 6, 135-144.
- Frymoyer, J.W. and Pope, M.H. (1983) Risk factors in low back pain. An epidemiological survey. *Journal of Bone and Joint Surgery* 65, 213-218.
- Frymoyer, J.W. and Cats-Varil, W.L. (1991) An overview of the incidences and costs of low back pain (review). *Orthop Clin Nor Am* 22, 263-271.
- Goel, V.K., Voo, L.M., Weinstein, J.N., Liu, Y.K., Okuma, T. and Njus, G.O. (1988) Response of the ligamentous lumbar spine to cyclic bending loads. *Spine* 13, 224-230.
- Granata, K.P. and Marras, W.S. (1995) The influence of trunk muscle coactivity on dynamic spinal loads. *Spine* 20, 913-919.
- Grant, K.A., Habes, P.J and Tepper, A.L. (1995) Work activities and musculoskeletal complaints amongst preschool workers. *Applied Ergonomics* 26, 405-410.
- Hansson, T.H., Keller, T.S. and Spengler, D.M. (1987) Mechanical behavior of the human lumbar spine.II. Fatigue strength during dynamic compressive loading. *Journal of Orthopaedic Research* 5, 479-487.
- Hedman, T.P. and Fernie, G.R. (1997) Mechanical response of the lumbar spine to seated postural loads, *Spine* 22, 734-743.
- Herrin, G.D., Jariedi, M. and Anderson, C.K. (1986) Prediction of overexertion injuries using biomechanical and psychophysical models. *American Industrial Hygiene Association Journal* 47, 322-330.
- Jäger, M. and Luttman, A. (1980) Biomechanic analysis and assessment of lumbar stress during load lifting using a dynamic 19-segment human model. *Ergonomics* 32, 93-112.
- Jäger, M., Jordan, C., Luttman, A. and Laurig, W. (2000) Evaluation and assessment of lumbar load during total shifts for occupational manual materials handling jobs within the Dortmund Lumbar Load Study-DOLLY. *International Journal of Industrial Ergonomics* 25, 553-571.
- Kelsey, J.L., Githens, P.B., White III, A.A, Holford, T.R., Walter, S.D., O'Connor, T., Ostfeld, A.M., Weil, U., Southwick, W.O. and Calogero, J.A. (1984) An epeidemiologic study of lifting and twisting on the job and risk for acute prolapsed lumbar intervertebral disc. *Journal of Orthopaedic Research* 2, 61-66.
- Kumar, S. (1990) Cumulative load as a risk factor for back pain. *Spine* 15, 1311-1315.

- LeVeau, B.L. (1991) *Biomechanics of Human Motion*, Williams, Marion.
- Marras, W.S. and Granata, K.P. (1997) Changes in trunk dynamics and spine loading during repeated trunk exertions. *Spine* 22, 2564-2570.
- Marras, W.S., Lavender, S.A., Leurgans, S.E., Rajulu, S.L., Allread, W.G., Fathallah, F.A. and Ferguson, S.A. (1993) The role of dynamic three dimensional trunk motion in occupationally-related low back disorders. *Spine* 18, 617-626.
- McGill, S.M. and Norman, R.W. (1985) Dynamically and statically determined low back moments during lifting. *Journal of Biomechanics* 18, 887-895.
- McGill, S.M. and Norman, R.W. (1986) Partitioning of the L4-L5 dynamic moment into disc, ligamentous, and muscular components during lifting. *Spine* 11, 666-677.
- McGill, S.M. (1992) A myoelectrically based dynamic 3D model to predict loads on the lumbar spine tissues. *Journal of Biomechanics* 25, 395-414.
- McGill, S.M. (1997) The biomechanics of low back injury implications on the current practice in industry and the clinic. *Journal of Biomechanics* 30, 465-475.
- Mientjes, M.I.V., Norman, R.W., Wells, R.P. and McGill, S.M. (1999) Assessment of an EMG based method for continuous estimates of low back compression during asymmetrical occupational tasks. *Ergonomics* 42, 868-879.
- Milburn, P.D. and Barratt, R.S. (1999) Lumbosacral loads in bed making. *Applied Ergonomics* 30, 263-273.
- Mital, A., Nicholson, A.S., Ayoub, M.M. (1993) *A Guide to Manual Materials Handling*. Taylor & Francis, Bristol, PA.
- Moore, J.S. and Garg, A. (1992) Ergonomics: low back pain, carpal tunnel syndrome, and upper extremity disorders in the workplace. *Occupational Medicine* 7, 609-620.
- Myklebust, J.B., Pinar, F., Yogandan, N., Cusick, J.F., Maiman, D., Myers, T.J. and Sances, A. (1986) Tensile strength of spinal ligaments. *Spine* 13, 526-530.
- Nachemson, A. (1975) Towards a better understanding of low back pain: A review of the mechanics of the lumbar disc. *Rheumatology and Rehabilitation* 14, 129-143.
- Neumann, P., Osvalder, A.L., and Nordwall, A. (1992) The mechanism of initial flexion distraction injury in the lumbar spine. *Spine* 17, 1083-1090.
- NIOSH (1981) *Work Practices Guide for Manual Lifting*, NIOSH Technical Report NO. 81-122, US Department of Health and Human Services, National Institute for Occupational Safety and Health, Cincinnati, OH.

- Norman, R., Wells, R., Neumann, P., Frank, J., Shannon, H., Kerr, M. and the Ontario Universities back Pain Study Group. (1998) A comparison of peak vs cumulative physical work exposure risk factors for the reporting of low back pain in the automotive industry. *Clinical Biomechanics* 13, 561-573.
- Potvin, J.R. and O'Brien, P.R. (1998) Fatigue related responses of the trunk muscles to a prolonged, isometric twist exertion. *Clinical Biomechanics* 12, 306-313.
- Osvalder, A.L., Neumann, P. and Nordwall, A. (1990) Ultimate strength of the lumbar spine in flexion an in vitro study. *Journal of Biomechanics* 23, 453-460.
- Osvalder, A.L., Neumann, P. and Nordwall, A. (1993) A method for studying the biomechanical load response of the in vitro lumbar spine under dynamic flexion shear loads. *Spine* 26, 1227-1236.
- Petrofsky, J.S. and Lind, A.R. (1978) Metabolic, cardiovascular and respiratory factors in the development of fatigue in lifting tasks. *Journal of Applied Physiology* 45, 64-68.
- Potvin, J.R., Norman, R. and Wells, R. (1990) A field method for continuous estimation of dynamic compressive forces on the L4/L5 disc during the performance of repetitive tasks. *Proceedings of the 23rd Annual Conference of the Human Factors Associations of Canada*. Ottawa: Human Factors Association of Canada.
- Potvin, J.R., McGill, S.M. and Norman, R.W. (1993) Quantification of erector spinae muscle fatigue during prolonged, dynamic lifting tasks. *European Journal of Applied Physiology* 67, 554-562.
- Potvin, J.R. and O'Brien, P.R. (1998) Trunk muscle co-contraction increase during fatiguing, isometric, lateral bend exertions. *Spine* 23, 774-781.
- Punnett, L., Fine, L.J., Keyserling, W.M., Herrin, G.D. and Chaffin, D.B. (1991) Back disorders and nonneutral trunk postures of automobile assembly workers. *Scandinavian Journal of Work and Environmental Health* 17, 337-346.
- Solomonow, M., Zhou, B-H., Baratta, R.V., Lu, Y. and Harris, M. (1999) Biomechanics of increased exposure to lumbar injury caused by cyclic loading: Part 1. Loss of reflexive muscular stabilization. *Spine* 24, 2426-2434.
- Sparto, J.P. and Parnianpour, M. (1998) Estimation of trunk muscle forces and spinal loads during fatiguing repetitive trunk exertions. *Spine* 23, 2563-2573.
- Timm, K.E. and Malone, T.R. (1989) Back injuries and rehabilitation. *Sports Injury Management* 2, 1-10.

Troup, J.D. (1976) Mechanical factors in spondylosolisthesis & Spondylolysis. *Clinical Orthopaedics* 117, 59-67.

Vanharanta, H. (1989) Etiology, epidemiology, and natural history of lumbar disc disease. *Spine* 3, 1-26.

Watkins, J. (1999) Structure and Function of the Musculoskeletal System, (1st ed. Edited by Merz Bott, S. and Mitchell, C.). Human Kinetics, Illinois.

Whiting, W.C and Zernicke, R.F. (1998) Biomechanics of Musculoskeletal Injury, (1st ed., edited by Robertson, L.). Human Kinetics, Illinois.

Yingling, V.R. and McGill, S.M. (1999) Anterior shear of spinal motion segments. Kinematics, kinetics and resultant injuries observed in a porcine model. *Spine* 24, 1882-1889.

Vita Auctoris

Name: Christa Lauder

Place of Birth: Edmonton Alberta, Canada

Year of Birth: 1976

Education: St. Josephs High School, Windsor, Ontario
1990-1995

University of Windsor, Windsor, Ontario
1995-1999 B.H.K